Cover picture:

The front cover of this report shows the HearLab, a computer-based audiological test suite being developed at NAL. It is a general-purpose interface unit for stimuli presentation and response acquisition, and is operated from the PC which controls stimulus levels, signal pathways and response conditioning. A battery of audiological tests will be developed for the HearLab as software modules. The first test module is for assessing acoustic evoked cortical electric potentials of babies and infants wearing hearing aids. More details are on pp 6-8.

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Research Director’s Overview

NAL had 45 projects active at various stages during the year, which is a higher-than-optimal number. Pleasingly, 16 projects reached completion during the year, allowing us to go into the new year with a more manageable number. A further three projects were terminated during the year. This report includes a representative sample of current and recently completed projects, each report giving enough detail to be of interest to clinicians and researchers in our field, but hopefully expressed in a manner that is accessible to the interested lay reader. In preparing this report, we are cognisant of not making this report the primary means of reporting project outcomes. The more widespread readership resulting from publication in leading journals, and the accompanying peer review of our work, remain the most important methods of dissemination of NAL research. Our web site, www.nal.gov.au, is an increasingly important means of dissemination, and presentations at conference remain an important avenue for dissemination, cross-fertilisation of ideas, and research staff training.

Hearing Assessment

Several years ago, we identified that there would be a growing need for methods to fit, fine-tune, and evaluate hearing aids provided to infants whose hearing loss is detected by a universal screening program within a few days of birth. With the good progress made to the HearLab general-purpose audiological test equipment during the year, we seem well-positioned to provide assistance to clinicians helping infants in a very concrete way over the next year or two. The hardware development of HearLab was completed, including the high-speed (USB) connection to any personal computer. The base software (firmware) that underlies all modules, and the software specific to the first module, both made steady progress. This equipment will be a major way in which novel tests originated by NAL find their way to clinicians for many years to come. The concept will go beyond even that, however. Even before HearLab’s release (anticipated to be in September 2005) one internationally leading researcher in the UK has agreed to have an innovative test procedure originating in his laboratory made available as a module within HearLab.

Certainly the first module to be released, which enables the effectiveness of hearing aids fitted to infants to be rapidly assessed via auditory evoked cortical potential, appears to be an extremely useful addition to the tools available to clinicians. Several articles in this Annual Report relate to various aspects of the Aided Cortical Assessment module for HearLab. The article on the HearLab hardware shows how the module will be able to allow for room acoustic characteristics to reliably reproduce test stimuli even under free-field conditions with loudspeakers of variable quality in a wide range of test rooms. Our continuing use of cortical assessments on infants fitted in nearby hearing centres is providing a steadily growing database of results. The accompanying research also includes results on children with normal hearing, which is enabling the maturational changes in early infancy to be better documented. It is not yet understood in what way acoustic stimuli must differ before the resulting cortical responses also differ. In past research we have shown that selected speech sounds that differ markedly in spectral shape and temporal characteristics reliably lead to differentiated cortical responses. One study outlined in this report extends that knowledge to a wider range of speech sounds. It is certainly now clear that many speech sounds which are easily differentiated behaviourally do not lead to differentiated cortical responses. A long-held aim of NAL has been to make novel tests and procedures available in a way that lets the maximum number of clinicians use them reliably. Accordingly, we are building a considerable level of automation into the cortical assessment module of Hearlab, to assist clinicians who are not specialists in electrophysiological measurement. An overview of the principles and assumptions behind the automatic
waveform analysis methods (based on MANOVA) we are using is contained in a further article. Although it is not a mainstream part of NAL work, an article in this report describes the use of cortical (discriminative) potentials to better understand the impact of auditory processing disorders on children with delayed reading ability.

It is already clear, from our research and others, that the cortical assessment test can evaluate the audibility of amplified speech, and that the test gives information about the normalcy of the auditory processing system once speech is amplified sufficiently to be audible. Our research into the use of cortical potentials to guide the detailed fine-tuning of hearing aids continues.

Of course, the availability of cortical assessment does not remove the need for behavioural assessment methods. Many paediatric clinicians believe there is value in having access to a variety of narrow-band stimuli to help achieve and maintain a child's interest. As shown in an article in this report, our first attempt to confirm this belief did not meet with success. Three of the four stimuli used were just as effective as the warble tones traditionally used, and the other was less effective. We will be re-examining this from another perspective next year.

There is an increasing need for efficient methods for estimating a hearing-impaired person's loss of ability to suppress background noise while understanding speech. During the year we devised a new test, the Beautifully Efficient Speech Test (BEST), that should enable testing to be done in a more efficient manner than would otherwise be the case. The test homogenises the difficulties of scored items in a test of sentence perception by varying the signal-to-noise ratio from one word to the next. Sentence scores vary from close to 0% correct up to close to 100% correct over very small range of overall signal-to-noise ratios. Pilot testing has shown that sentence intelligibility changes with signal-to-noise ratio at a very high rate of 34% per dB. This enables the signal-to-noise ratio for 50% correct to be found with high accuracy in a very short time. When the research is completed, and the test made available, we expect that clinicians will want to use it to assess the level of need for individual clients to have directional microphone technology, and the degree of improvement that particular devices offer in the situations tested.

**Hearing loss prevention**

NAL is fortunate to have available test results for over 100 types of hearing protector, each measured on 20 or so subjects. A paper in this report shows that the attenuation rating a protector receives depends strongly on the range of performance measured across people. Low-attenuation devices receive a low rating not because everyone gets a low degree of protection, but because some people get an extremely low degree of protection, while some get a high degree of protection. This spread of performance is doubly bad: some people get less attenuation that they anticipated, and perhaps get insufficient attenuation to protect their hearing. Other people get more attenuation than they anticipated, as the protector's rating is heavily influenced by the poor scores. Excessive attenuation can cause people to discontinue use, and so these people too then expose their hearing to damage.

Our previous research (see 2002/2003 report) illustrated the difficulty of delivering messages that actually alter people's behaviour in health-safety issues. A suggestive, but non-significant trend to emerge from the study was that people who were given positive messages about their hearing (i.e. how well preserved it was) were more likely to take steps to protect their hearing than those who were advised that some damage had already accumulated. NAL will be engaging in considerably more research of this type in an effort to discover how to best motivate people to protect their hearing in whatever ways are available to them. As an aid to the delivery of hearing protection messages, we produced on CD a simulation of hearing loss for several degrees of hearing loss. We hope that the demonstration of the differences between listening with normal hearing and listening with a sensorineural hearing loss will provide the motivation for those with normal hearing to keep it that way. A future interactive version, that can simulate any desired loss and process any desired source material, will be constructed as a HearLab module.

**Hearing rehabilitation devices**

Research sometimes has to be done even though the outcomes can confidently be predicted. One such case is the study on linear versus non-linear amplification in children included in this report. Although it is now often the case that clinicians prescribe non-linear amplification for children, and we have recommended it for some years on theoretical grounds, the practice is still far from universal.
We were aware that there is no convincing research evidence supporting, and indeed, necessitating the practice. The study report here was done in a controlled, double-blind manner, with due allowance for acclimatization effects. It shows that non-linear amplification is convincingly superior in terms of speech perception at low input levels, localization accuracy at low and medium input levels, and functional performance in real life.

Localization research remains an important part of NAL’s work, enabled by our several high quality anechoic chambers. A paper in this report shows that modern high-performance hearing aids can unwittingly decrease localization accuracy when the hearing aid in one ear automatically adapts to a directional pattern that is different from the hearing aid in the opposite ear. Irrespective of the signal processing options in the hearing aids, confusion between frontal and backward source locations in the horizontal plane (and presumably between all other source locations with the same azimuth but different elevations) remains a major source of confusion. All these results have strong implications for future hearing aid design.

The advent of programmable and fully digital hearing aids has had implications for battery performance. Whereas once it was acceptable for a hearing aid battery to momentarily reduce its output voltage under high load conditions, modern hearing aids are likely to change programs, or otherwise alter the amplification characteristics when such interruptions occur to the power supply. A report in this study examines the implications for how batteries need to be specified and measured to avoid such disruptions to use.

**Hearing rehabilitation procedures**

Although NAL is well known for its prescription procedures based on pure-tone hearing thresholds, we have never taken the view that the resulting prescriptions are necessarily optimal for any individual client. Indeed, we have used various fine-tuning methods in research studies and have published on methods that clinicians can use to systematically fine-tune hearing aids. A development that we are very excited about is the Trainable Hearing aid, which we have developed in conjunction with Melbourne University and the Bionic Ear Institute as part of the Cooperative Research Centre for Cochlear Implant and Hearing Aid Innovation. This scheme allows clients to fine-tune their own hearing aids, in their own environments, in their own time. Initial results have been extremely promising, and this report contains an article that examines the accuracy with which clients can make adjustments to the response of a hearing aid with different combinations of controllers.

We continue to strive to better understand how well hearing-impaired people can extract information from speech once it has been amplified. Two articles outline a very large study that we have almost completed, with the aim of finding whether there is anything that can be measured on a client that can lead to better predictions of speech intelligibility than is possible based on pure tone thresholds alone. So far, the only unequivocal answer is age! A proper understanding of these results is critical to deducing optimal hearing aid prescriptions, and will directly affect the forthcoming NAL-NL2 prescription.

**Contract research**

Several contract research projects were again commenced or completed during the year. One of these projects has already been released to the public by its sponsor, and provided a very interesting useful insight into the effects of advanced hearing aid signal processing on the wearer’s ability to localise sounds.

**Overview of NAL work**

The Research Committee and the Human Research Ethics Committee continued to give great service to NAL through their expertise and diligence. Without their generously donated time, NAL would not be able to perform research. We are particularly grateful to Dr Keith Joseph who continues to chair the Ethics Committee and to Professor Field Rickards who continues to chairs the Research Committee and who also serves on the Ethics Committee.

**HARVEY DILLON**
Computer-Based Audiological Test Suite: HearLab

Investigators: Teck Loi, Barry Clinch, Dan Zhou, Isabella Tan

This project is being carried out in collaboration with CRC Hear.

Background: NAL embarked on a project aimed at overcoming barriers in the path of bringing new audiological tests to clinicians. Typically new tests require new instruments to be developed at great cost before they are suitable for clinical use. This leads to considerable time lag between research findings and their application in clinical practice. The high cost of developing new instruments can prohibit or delay the take-up of new tests. NAL began development of a versatile instrument, HearLab, based on the standard PC. The approach was to develop a general-purpose interface unit for stimuli presentation and response acquisition. This interface unit allows the PC to fully control stimulus levels, signal pathways and response conditioning. With this approach new audiological tests can be implemented entirely by new software modules. At minimal production cost, the power of modern PCs can be utilised to generate any novel stimuli and to perform complex signal processing and analysis on the responses acquired. Traditionally expensive front panels are eliminated by a software user interface. Thus expensive and complex research equipment used to develop the new tests can be transferred into a clinical setting at an affordable cost. Time-wise, it is possible that as soon as the research is completed, the new test can be applied in the clinics by the release and installation of a new software module for HearLab.

Aided Cortical Analyser (ACA): NAL research studies have established that analysing acoustic evoked cortical electric potentials (AECEP) offers an assessment of hearing aid fitting in infants. AECEP is best recorded while the subject is awake. This is advantageous in clinical practice as sedation is avoided. The equipment used for this area of research has been expensive and cumbersome for widespread clinical use. Stimulus presentation, response acquisition and processing and results to be displayed are complex and multidimensional, as are the test protocols. Statistical computation for response detection and differentiation has been developed during research to assist assessment. To integrate all the required functions in a clinical instrument would be desirable. The first module of the HearLab test suite to be made available will therefore be the ACA for the assessment of hearing aid fittings in infants. With implementation of universal screening programs in many countries, fitting of hearing aids in infants as early as 4-8 weeks is becoming more common. At this young age, it is difficult to assess hearing aid effectiveness using behavioural assessment techniques. NAL’s research provides an objective assessment of hearing aid fitting. The demand for cortical testing at an affordable price is already arising. Making ACA simple to use and set up is also a challenge. The following pictures illustrate that the HearLab equipment for cortical assessment is practical for clinical setups.

Front and back view of the main interface unit used for all HearLab modules. Additional sockets on the back panel are for connecting other input and output devices for future modules. It has built-in amplifiers to drive loudspeakers to required acoustic levels at the test position. For ACA, a monitoring microphone may be connected to it and can be used to automatically compensate for room acoustics and to automatically calibrate stimulus levels.
**Automated acoustic equalisation:** Calibration of tone-bursts is straightforward. Stimulus levels at audiometric frequencies are measured at the subject location and level compensation is achieved with automatic adjustment of a gain control. ACA also uses short duration speech phonemes as acoustic stimuli presented by free-field loudspeakers in a test room. In a clinical setup, the test room acoustics, the positioning of loudspeakers and the location of test subject are all variables that affect the actual stimulus level and its spectrum. Calibrating the stimuli to specified levels and maintaining a correct spectrum would traditionally require equipment such as signal generators, microphones and spectrum analysers to arrive at a setting of equalisers inserted in the playback chain. The process needs to be repeated whenever the loudspeaker or physical arrangements are altered. Only well-resourced clinics or research laboratories can manage calibration and acoustic equalisation in the complex set up. To overcome this, HearLab will contain software that automates level setting and spectrum compensation, without additional equipment. The clinician can initiate the calibration procedure at any time or as prompted by ACA at regular intervals.

Acoustic level against frequency obtained with warble tone in a typical set up in NAL Cortical Test room (3.9m X 4.0m X 2.4m, front loudspeaker and test position distance of 2.0m)

For speech stimuli, Hearlab employs digital signal processing to the speech stimulus wavefile to compensate for loudspeaker and room acoustics. The aim is to ensure that speech amplitude and timing are adequately presented acoustically to the subject irrespective of the stimulus playback set up. By measuring the acoustic response playback path at the test position, HearLab can estimate the parameters of the filters that have the reverse transfer function of the playback path. The result is illustrated by the match between the power spectrum

Statistical measures indicating the significance level of response detection and response differentiation are displayed in addition to grand averaged responses and ongoing EEG traces.

Connector box for electrophysiological modules. Specially developed prototype active electrodes are shown with the connector box which can be located close to the subject.
of wavefile ‘Tae’ and the power spectrum of adjusted ‘Tae’ recorded at the test position.

Prototype Development: The interface unit and the active electrodes are fully functional. Three units have been constructed. These prototypes are employed by the software group in testing the ACA software under development. The aim over the next 12 months is to complete instrument prototypes and carry out clinical trials.

Significance: This project provides a vehicle to widen and accelerate the take-up of NAL’s research findings, to the benefit of hearing-impaired people. Revenue generated will be used to fund further research and development. NAL/CRC are currently holding discussions with a commercial collaborator concerning marketing, manufacture, distribution, sales and support of HearLab. Other researchers in the field may wish to make their results more readily available to clinicians via implementation within HearLab.

Maturational changes in the obligatory cortical evoked response

Investigators: Maryanne Golding, Suzanne Purdy, Harvey Dillon, Mridula Sharma, Katrina Agung

Background: While auditory responses recorded from the brainstem mature over the first 18 months of life\(^1\), cortical responses have a much longer developmental time course. As infants with normal hearing mature, cortical responses change significantly with respect to the shape and latency of the major components over the first 14-16 years of life\(^2\,\,4\).

The newborn infant cortical response in response to speech stimuli is dominated by a series of positive peaks with a prominent peak at 200 to 300 msec when recorded at the midline\(^5\,\,7\). In childhood, the morphology is dominated by a large positivity with an approximate latency of 100 msec followed by a negativity (175 to 275 msec) and a second positivity (300 msec)\(^8\,\,9\). By adult years (i.e., over 20 years), the dominant component is a negativity (80 – 120 msec) that is preceded and followed by positive components (i.e., P1 at 50 to 70 msec, and P2 at 150 – 200 msec). While the child's negative component is often labelled as N1 it is likely to have a different cortical origin to that of the adult N1 and is therefore not the same response\(^8\). Similarly, the infant single component is often labelled as P1 but could be expected to have different generators to that of the adult P1. These morphological changes with age are likely to reflect underlying developmental changes in the response generators such as improved synaptic efficiency arising from increased axon myelination and maturation of intra and inter-hemispheric connections throughout the cortex\(^10\,\,11\).

Several studies have examined this latency change as a function of age particularly in normal hearing children and adults or children with cochlear
implants but few detail the changes in latency over the first few months of life for normal hearing infants and fewer still examine latency changes in children who wear hearing aids\textsuperscript{4,5,8-10}.

Drawing from a series of studies on cortical evoked responses conducted at NAL, the purpose of this report was to examine the latency shift for major response components in normal hearing infants and adults. While variations in latency could be expected to arise from differences in the speech stimulus used to elicit the response (e.g., speech segments with differing voice-onset-cues\textsuperscript{7}), the differences are subtle enough to combine into single graphs for this purpose.

\textbf{Method}

\textbf{Subjects:} There were 54 infants aged 0.2 to 0.75 years (mean 0.42 yrs, SD 0.12) with normal tympanometry, OAEs and apparently normal hearing. Adults (\(N = 22\)) had a mean age of 31.4 years (SD 10.86) and pure tone thresholds \(\leq 25\) dB HL from 250 Hz to 4000 Hz in both ears.

\textbf{Procedure:} Responses were recorded using the Neuroscan™ system with electrodes positioned at either Cz alone or at Cz, C3 and C4 referenced to the right mastoid with forehead as ground. Adult subjects also had an electrode positioned above the eye to monitor eye-blinks. During cortical testing, adults were seated comfortably with their left ear plugged and watched sub-titled DVDs. Infants listened binaurally, were awake and seated on their mother’s lap, distracted by another adult if required. Stimuli were delivered via a loudspeaker at 65 dB SPL positioned on the right hand side at 45 degrees azimuth, 1 meter from the head. Individual sweeps of the electroencephalic (EEG) activity were amplified and analog filtered on-line at 0.1 – 100 Hz using a 24 dB/octave slope, and subsequently filtered off-line at 1-30 Hz. The recording window consisted of a 100 msec pre-stimulus baseline and a further 600 msec.

A number of different speech segments were used across the individual experiments from which this data is drawn, but not all subjects were assessed with all stimuli. Each stimulus was presented until 100 responses were accepted. Each stimulus block was presented where possible on two occasions, and the order of the stimulus blocks was randomised.

\textbf{Results:} Infant responses were divided into two groups: those with a single positive peak (P1) and those with two positive peaks (P1 and P2). Where peaks could not unambiguously be located, they were identified based on the presence of local minima that preceded and followed the positivity. Visual lines of “best-fit” were applied either side of the peak, beginning with the local minima and ending at the nearest local maximum within the broad peak region. The point of intersection of the two lines indicated the latency. P2 was only recorded when the second peak had a latency at least 80 msec longer than that of the first peak to ensure that they were separate peaks.

Adult responses were combined and reported as a single reference group centred at the mean age. Infant and adult results were reported for Cz only with the response latency to all stimuli identified and averaged to provide a single data point for each of P1 and P2 (where present).

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure1.png}
\caption{Infants with one positive peak (P1) only and adult P1,N1,P2 response latencies (infants, N=25; adults P1 (N=21),N1/P2 (N=22)).}
\end{figure}
**Figure 1** shows the latency for infant responses with a single positive peak and the adult P1, N1 and P2 response as a function of age. There appears to be significant variability in latency for the infant response with a mean of 205 msec (SD 25). This infant P1 latency is longer than the mean adult P1 latency of 65 msec (SD 7) but similar to the adult P2 response latency (N=22) with a mean of 199 msec (SD 19).

**Figure 2** shows the response latencies for P1 and P2 in infants and adults with two positive peaks. There is a decrease in latency for both P1 and P2 as a function of age with the mean P1 latency being 186 msec (SD 28) for infants and 65 msec (SD 7) in adults, while the mean P2 latency is 333 msec (SD 28) for infants and 199 msec (SD 20) in adults.

The mean age of infants with a single positive peak was 0.39 years (SD 0.11) while the mean age of infants with two positive peaks was 0.46 years (SD 0.12). An independent t-test showed a significant difference in age (t (52) = -2.107; p = 0.04) with infants who have two positive peaks being older than those with a single peak. An independent t-test on the latency of P1 in infants with a single peak compared with those who have two positive peaks, showed that the P1 latency was significantly shorter for those with two positive peaks (t (52) = 2.72; p = 0.009). For P1 (all infants combined), the linear regression equation of latency with age was significant (p < 0.001) with a slope of -11 msec/month.

**Discussion:** The development of cortical responses with age is far more complex than responses arising from the brainstem, which should not be surprising given the extended time course required for complete maturation of the cortex. The metamorphosis of the infant response into the adult P1/N1/P2 response is not well characterized and made far more complex in the literature by inconsistency in nomenclature, as well as differences across studies with respect to age ranges, recording sites and stimuli.

The cortical generators of these maturational changes are complex. There is some evidence for differences in maturation time course for differing stimuli (e.g., clicks versus speech stimuli) suggesting later maturation and differing neural substrates required for processing speech versus non-speech stimuli\(^\text{12}\), as well as differences in neural generators across age groups that may be explained by the differing developmental time course for the layers within the neocortex. It appears that layer I axons mature within the first few months of life, but lower layer III, IV, V and VI are not mature until 3–5 years of age, while layers II and upper layer III are finally mature around 5–12 years of age\(^\text{13}\). In adults, positive cortical evoked response components, such as P1, are likely to arise from thalamic input to pyramidal cells in lower layer III and IV\(^\text{11}\) but if this region is not mature until 3-5 years of age, it seems feasible that the infant positivity (P1) has a different origin. Similarly, the developing N1 in children appears to have a deep layer III origin but is more superficially generated in adults\(^\text{8}\).

Our results show decreasing P1 latency with increasing age in normal hearing infants which is consistent with reports in the literature although maturational studies of infant responses are rare\(^\text{5,8,10}\).
Our results also suggest that many infants are likely to develop two positive peaks within the first 12 months of age and that both these peaks continue to decrease in latency as the infant matures. Latency/age data will continue to be collected for normal hearing infants and similar data will be examined in infants and children with hearing loss. This data will be added to the Aided Cortical Assessment module within HEARLAb, enabling the clinician to compare their patients’ results with those of normal maturational processes for normal-hearing children and those with sensory loss.

References

for the CAEP post-fitting, together with the observed improvements in attentiveness and babbling in the infants, confirmed the usefulness of this technique in assuring the effectiveness of habilitation. Kurtzberg (1989) also showed that CAEPs were useful in measuring the efficacy of amplification. She reported that a young child with moderate-severe hearing loss had a poor discriminative CAEP to speech segments presented at 90 dB peSPL but with hearing aids fitted, a large robust response was observed which indicated that her auditory system was able to discriminate acoustic features.

**Previous research at NAL:** Using the Neuroscan™ system at NAL, Purdy et al (unpublished data) carried out a study to establish normative cortical evoked potential data for adults (n=10) and infants (n=20) using 3 speech sounds, /m/, /g/ and /t/, which have emphasis in the low, mid and high frequency regions respectively. Data from these adults and infants indicated that cortical potentials evoked by different stimuli could be reliably obtained and could be distinguished within individual subjects based on the morphology, amplitudes and latencies of the responses.

A later experiment examined the sensitivity of CAEPs in forty hearing-impaired children with moderate – profound hearing loss. Results indicated that an aided cortical response to 65 dB SPL could be elicited for all children with a moderate hearing loss, for most children with severe and for half of the children with profound hearing loss. In some cases, these responses were seen after some adjustment to the hearing aid fitting.

Purdy et al. then assessed ten children aged approximately 6-12 years, with stimuli which simulated hearing aid fine tuning adjustments. This showed that CAEPs were sensitive to changes of 6 dB per octave.

As a result of these experimental findings, a long-term clinical study was devised where the families of newly diagnosed infants or children who were difficult to assess behaviourally when aided, were asked by their family audiologist if they would participate in research into the usefulness of the technique in clinical practice.

As part of this research, the clinical protocols required for the efficient use of the procedure in an audiology clinic will be devised and these findings will be incorporated into the first module of HEARLab (see article in this publication).

**Clinical Study Methodology:** Brain electrical activity was recorded using the Neuroscan™ system with electrodes positioned at Cz, C3 and C4 referenced to the right mastoid with forehead as ground. During cortical testing, infants were awake and seated on their mother’s lap, distracted by another adult if required. Stimuli were delivered via a loudspeaker positioned at 45 degrees azimuth on the side of the subject closest to the ear being evaluated. Individual sweeps of the electroencephalic (EEG) activity were amplified and analog filtered on-line at 0.1 – 100 Hz using a 24 dB/octave slope, and subsequently filtered off-line at 1-30 Hz. The recording window consisted of a 100 msec pre-stimulus baseline and a further 600 msec.

Each stimulus was presented until 100 responses were accepted. Each of these stimulus blocks were presented where possible on two occasions in random order. Stimuli were delivered at 65 dB SPL, or greater if no response was seen, up to a maximum of 85 dB SPL. The stimuli were three speech sounds, /m/, /g/ and /t/ with an inter-stimulus interval of 1125ms

**Case Report:** S8 was born at full term with a syndrome that is often related to a permanent mixed hearing loss. He received a “refer” result from two automated auditory brainstem response (AABR) screening tests performed as part of the NSW infant hearing screening program (SWISH). He was referred for diagnostic audiology at 2.5
weeks of age and the resulting auditory brainstem response (ABR) thresholds were as follows:

<table>
<thead>
<tr>
<th></th>
<th>500Hz</th>
<th>1000Hz</th>
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<tr>
<td>Right ear dB nHL</td>
<td>90NR</td>
<td>90</td>
<td>80</td>
<td>60</td>
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<tr>
<td>Left ear dB nHL</td>
<td>90</td>
<td>90</td>
<td>70</td>
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Distortion-product otoacoustic emission testing, which measures outer hair cell function, was also performed and emissions were absent in both ears. Impedance audiometry indicated a normal tympanogram for the left ear and some negative pressure for the right ear.

Fully digital hearing aids with two-channel input compression were fitted using the results of the tone burst ABR and the NAL-NL1 prescription procedure as per the Australian Hearing protocols for children.

S8’s parents agreed to participate in our research using CAEPs so that the hearing aid fitting could be optimized as early as possible. The first assessment occurred at approximately 4 months of age with Figures 1 to 6 showing the grand average CAEP results (recorded at Cz) for the three speech stimuli. On the basis of the complete set of test results, recommendations were then made to the family’s paediatric audiologist to increase the low and mid frequency gain in the right ear, and to increase the low and high frequency gain in the left ear.

Following the adjustment of the hearing aids and a short period (4 weeks) to allow for acclimatisation, S8 was reassessed. CAEP results showed clearer responses to many speech stimuli as shown in Figures 1 to 6, which suggested that the adjustments made to the hearing aids improved the child’s detection of speech sounds at levels approximating average speech.
Discussion: S8 was diagnosed with a severe hearing loss following referral from the NSW newborn hearing screening program (SWISH). He was subsequently fitted with hearing aids in keeping with the Australian Hearing protocols for children.

The availability of CAEP results from this child’s participation in the research allowed the family and the paediatric audiologist to fine-tune the hearing aids to ensure detectability of speech sounds earlier than would have been possible if waiting for behavioural thresholds to be obtained. The CAEPs provided objective evidence of the child’s ability to detect low, mid and high-frequency speech sounds at conversational level through his hearing aids, which is important information for his parents and teachers as they conduct auditory habilitation.

Evidence to date suggests that recording CAEPs is valuable in evaluating amplification in children who are unable to perform visual reinforcement orientation audiometry (VROA) either because of age or developmental delays. Infants will continue to be recruited to the clinical study underway at NAL, and results of electrophysiology, functional assessments and behavioural assessments will be entered into our database for analysis later next year.

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Objective verification of speech perception using cortical auditory evoked potentials

Investigators: Katrina B Agung\textsuperscript{1,2}, Suzanne C Purdy\textsuperscript{1,2,3}, Catherine McMahon\textsuperscript{2}, Harvey Dillon\textsuperscript{1}, Richard Katsch\textsuperscript{1}, Philip Newall\textsuperscript{2}

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Cortical auditory evoked potentials (CAEPs) may provide an objective electrophysiological method of validating hearing aid fittings in infants that are too young to participate in behavioural tasks. In recent years there has been an increase of interest in using CAEPs to assess speech perception (e.g. Ostroff et al., 1998; Tremblay et al., 2003). CAEPs are known to correlate closely with perceptual auditory threshold (Hyde, 1997). Further, CAEPs elicited by speech sounds, such as consonant-vowel syllables provide an opportunity to assess the functional integrity of auditory pathways involved in the acoustic analysis of speech (Sharma et al., 1997) that are important for the development of speech and language (Novak et al., 1989). Purdy et al., (2001a,b) found that cortical responses could be recorded to speech stimuli with low-frequency versus high-frequency dominant spectral energy ([mae] vs [tae]) to objectively demonstrate perception of phonemic differences, important for speech and language development in normally-hearing and hearing-impaired infants with well-fitted hearing aids.

The aim of the current study was to determine whether CAEPs elicited by a wide range of speech sounds could be used as an objective test to evaluate perception of speech sounds in infants. The Ling six sounds were chosen because the ability to discriminate these sounds behaviourally is thought to indicate a child's ability to discriminate speech, as these sounds have concentrations of energy across the entire range of speech frequencies (Ling, 2002). The Ling six sounds comprise the vowels: “a” as in car, “oo” as in two, and “ee” as in she, along with the phonemes: /s/ as in us, /sh/ as in fish, and /m/ as in me; The phonemes “or” as in hoard and “uu” as in hood were added for the Australian population (see Agung et al., submitted). The Ling speech sounds are often used by audiologists as a practical, quick behavioural test to assess which speech sounds an older hearing-impaired child and adult can identify and discriminate with amplification (Ling, 2002).
**Research Questions**

The following research questions were asked:

1. Can CAEPs be recorded in normally hearing awake infants and adults, to a range of supra-threshold speech sounds that encompass the speech frequencies?
2. Do CAEPs show significant differences in cortical response waveshape across speech stimuli?
3. Which stimulus duration is optimal for speech-evoked CAEP recording?

**Procedure:** CAEPs were recorded from 10 adults and 20 infants that were normally hearing. The Neuroscan STIM and SCAN (version 4.2) evoked potential system was used for stimulus generation and CAEP recording. Stimuli were presented bilaterally at a conversational level via a loudspeaker. Electrodes were placed at Cz (vertex) and A2 (right earlobe, reference) for adults and over the right mastoid in infants, with a ground electrode on the forehead. The inter-stimulus interval was 1125 ms. Adult participants were seated comfortably in a recliner chair in a sound treated booth. To keep alert during testing, participants watched a self-chosen movie with subtitles but without sound. Infant participants were seated on the lap of their mother or father and were kept alert during testing by being entertained with toys. Two stimulus durations of 500 ms and 100 ms (+/- 5ms) were used for each stimulus for adult participants. One stimulus duration of 100 ms was used for infant participants.

CAEP peak amplitudes and latencies were identified for each participant by two independent observers. Separate repeated measures analyses of variance (ANOVA) were performed, with stimulus as the repeated measures factor, to determine the effects of stimulus type and duration on the latencies and amplitudes of each peak (P1, N1, and P2). To determine whether these differences in the cortical response when elicited by the different speech sounds occurred for an individual participant a multivariate analysis of variance (MANOVA) was used.

**Findings:**

1. Cortical responses were elicited by all speech sounds for all adult and infant participants. This demonstrates that CAEPs can be reliably evoked by sounds that encompass the entire speech frequency range.

2. ANOVA results for the adult participants showed that speech sounds that were dominated by high frequency energy produced CAEPs that were significantly different in response amplitude when compared to sounds dominated by lower frequency energy. This reflects the tonotopic organization of the auditory cortex. Cortical areas that respond to low frequency auditory information are located more superficially (i.e. closer to the surface of the scalp) than cortical regions for higher frequencies (Jacobson et al., 1992). Low frequency speech sounds may therefore activate more superficial cortical regions and therefore produce larger amplitude cortical responses than higher frequency speech sounds, when surface scalp recording electrodes are used.

MANOVA performed for infant participants revealed that 15 of the 20 participants had significantly different cortical wave-shapes when comparing the lower-frequency vowel “oo” versus the higher-frequency sound “sh”. For the other stimulus pairs, there were between 3 and 13 subjects for whom the response to the two stimuli in the pair differed.

3. As shown in Figure 2 the amplitude of the CAEP was reduced when elicited by longer duration sounds compared to shorter duration sounds. While it is currently not clear why this occurs, it is known that the N1 peak of the CAEP is susceptible to habituation with changes in the inter-stimulus interval and stimulus repetition (Budd et al, 1998). This may also be the mechanism with increases in stimulus duration.
Figure 2. Grand average CAEP waveforms in adult participants for all speech stimuli. N1 and P2 peaks were significantly larger in amplitude and earlier in latency for cortical responses elicited by shorter (100 ms) duration stimuli (left) compared to longer (500 ms) duration stimuli (right).

**Conclusion:** CAEPs may be used to objectively measure perception of spectrally different speech sounds. Future research should be directed towards exploring how other features of speech, such as temporal characteristics, also influence the cortical response.

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**The detection and differentiation of CAEPs using a MANOVA procedure**

**Investigators:** Richard Katsch, Harvey Dillon

This project is carried out as part of CRC HEAR.

**Background:** Evoked cortical potentials can vary in morphology from person to person. Reliable automatic detection and differentiation of Cortical Auditory Evoked Potentials (CAEPs) are therefore essential if cortical assessment of aided functioning is to gain widespread use. Conventional existing techniques use a process of selecting sections of electro-encephalographic (EEG) data, which are time locked to an ongoing stimulus. These sections, referred to as epochs, are then combined to provide an average waveform. Details of the average waveform, such as the presence and time latency of various peaks, provide useful diagnostic
information. Such an average waveform analysis can require an experienced interpreter and considerable measurement and interpretation time. Useful data may also be lost in the process of constructing the averages; however the inspection of several hundred raw epochs is not practical as the elicited response is significantly contaminated with noise. As a compromise, the average of the even and odd waveforms are often displayed, but considerable judgement is needed on the part of the clinician to judge whether a waveform is present, given the similarity of the even and odd averages.

These considerations have led us to investigate whether the raw epoch data can be analysed statistically to provide a reliable answer to the question of whether a response is present for a particular stimulus, and whether responses to different stimuli are, in themselves, different.

Objectives: To develop a statistical procedure that can detect the presence of, and differentiate between CAEPs produced by different auditory stimuli. This procedure should be capable of being incorporated into the NAL HearLab ACA module.

Procedure: The procedure was developed within the MATLAB programming environment, using data that had been recorded with a commercial NeuroScan system. The data used was in the form of ‘.eeg’ files, which contain a sequence of recorded epochs, together with descriptive information. Each epoch within the series is a snapshot of the ongoing EEG activity that is time locked to the stimulus onset. This recorded data can be presented to the analysis programs in the same epoch-by-epoch way that will occur in the HearLab system.

A brief description of the process is:

- An ‘.eeg’ file is produced which contains epochs of Gaussian White Noise. This is the null, or no response, condition.
- For each stimulus response and the null condition:
  - Each epoch, typically of ~700 points, is processed separately to produce a vector containing a smaller number of variables – a “reduced epoch”.
  - The sequence of “reduced epochs” is regarded as a series of independent measures of their constituent variables (under the null hypotheses that no response waveform is present or that the responses to two different acoustic stimuli are identical).
  - The sequence of “reduced epochs”, together with a classification variable that indicates the stimulus presented, is used as an input to a Multidimensional Analysis of Variance (MANOVA) procedure.

- The results of the MANOVA procedure using the response to two different stimuli responses identifies whether the two stimuli reliably give rise to different cortical responses, presumably arising from activity in different cortical locations, and/or at different times. We refer to this as the differentiation ability of the procedure.
- The result of the MANOVA procedure using a stimulus response and the null response indicates whether the response is significantly different from random noise. We refer to this as the detection ability of the procedure.

The MANOVA procedure described here uses the processed voltage measurements within each of the epochs as a separate variable. As described above, this provides many different variables potentially useful in distinguishing the difference between the responses. Each variable is measured, as many times as there are epochs for each stimulus. Thus, we have more information here than a simple average of the responses; we also have information about the distribution of the measures of each variable. The MANOVA procedure makes assumptions about the nature of the measures, which are addressed below:
• The populations for each group are normally distributed. The variance-covariance matrix is the same for each population or group (i.e. stimulus in our case). All observations are mutually independent.

Figure 1 above shows that the distribution of a typical processed variable is very close to normal. This is to be expected given the large amount of random noise in each variable and the processing of multiple points in the "reduced epoch" process.

The variance-covariance matrix is the same for each population or group (i.e. stimulus in our case). The variance of each individual variable is greatly influenced by external noise, which we expect to have the same magnitude for both stimuli, and is the reason for the large degree of signal averaging normally used by conventional systems. The correlation between adjacent variables is increased by the low pass filtering which is part of the processing procedure. With the variance and the correlation established as similar between measures the variance-covariance matrix is determined and so this assumption seems reasonable.

• All observations are mutually independent.

As the epochs are normally separated by a resting period between stimulations it seems reasonable that correlations between variables in adjacent epochs will be small, under the null hypotheses of no response, or no difference between responses to different stimuli.

Conclusion: The procedure was shown to be capable of measuring significant differences between the responses to such stimuli as /m/ and /t/ in normal hearing children. A demonstration was also produced of a running significance measure, showing the variation of the discrimination and detection ability as the number of epochs processed increased. This type of display allows the clinician to make more rapid assessments of the presence of discrimination/detection ability in a subject.

Figure 2 above shows the evolution of the ‘p’ significance level with the number of epochs tested. The light and dashed traces are measures of the detection capability of the process and show how both responses have achieved a ‘p’ value of < .05 after about 50 epochs. The heavy trace is a plot of the ‘p’ value of the difference between the two stimuli. In summary: after 50 epochs both stimuli were significantly different from a ‘null’ condition and both were different to each other. The horizontal line marks the 0.05 significance level. This analysis makes no allowance, however, for the effect on true probability of performing multiple MANOVAs on gradually accumulating data.

Further work: A comparison will be carried out between the performance of the MANOVA-based procedure and the judgement of experienced human clinicians, when judging the presence of a response or the differentiation of the response to two stimuli.
Discriminative cortical auditory-evoked potentials for the diagnosis of auditory processing disorders in children

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Background: Definition, diagnosis and management of auditory processing disorders have been difficult issues that both parents and audiologists have wrestled with for some time. Several “consensus” conferences have tried to put together theoretical frameworks to assist professionals dealing with auditory processing disorders (APD). Jerger and Musiek (2000) reported results from the Bruton, USA conference. They concluded that “APD in its pure form is a deficit in the processing of information that is specific to the auditory modality, that may be exacerbated in unfavourable acoustic environments, and that may be associated with difficulties in listening, speech understanding, language development and learning”. Both the ASHA (American Speech-Language-Hearing Association, 1996) and Bruton (Jerger and Musiek, 2000) conferences produced recommendations for screening and diagnosis of APD using a battery of tests (behavioural, electroacoustical and electrophysiological).

Behavioural test batteries are readily available clinically and have withstood the test of time. Several clinical APD tests have been used for many years in different clinical populations, and with different age groups to diagnose APD. The disadvantage of behavioural test batteries is that the processes being tested are not just auditory processes but also non-auditory processes such as memory, attention, motivation and linguistic output, amongst others. Thus, poor results on behavioural APD tests may not always be due to APD. Hence electrophysiological tests are of much interest since the influences of non-auditory influences of attention, linguistic output and motivation can be reduced.

The purpose of this research was to determine the usefulness of electrophysiological techniques for diagnosing APD in children with reading disorder. Electrophysiological techniques measure electrical brain activity time locked to an external event (Steinschneider et al., 1992). Electrophysiological responses include obligatory cortical auditory evoked potentials, which are elicited by the presence of a sound, and discriminative cortical auditory evoked potentials, which occur when there is a change in a repetitive sequence of auditory stimuli. The most common discriminative cortical potential that has been used to assess auditory processing in different clinical populations is the mismatch negativity (Kraus, McGee, Carrell, Zecker, Nicol, & Koch, 1996; Schülte-Körne, Deimel, Bartling, & Remschmidt, 1998, 1999a, b; Baldeweg, Richardson, Watkins, Foale, & Gruzelier, 1999; Cunningham, Watkins, Foale, & Gruzelier, 1999; Cunningham, Nicol, Zecker, & Kraus, 2000). Mismatch negativity (MMN) is an objective electrophysiological response that closely matches behavioural measures of auditory discrimination. MMN has the advantage over behavioural measures that the subject does not need to pay active attention to the stimuli and could be doing something entirely different during the course of testing, such as watching a silent movie or reading a book. A range of stimuli was used to elicit MMN including pure tones, chords (complex tones) and speech stimuli.

Research Questions
1. Do children with reading disorder have auditory processing deficits?
2. If children with reading disorder have APD, is there a specific APD pattern?
3. Are there any changes in auditory processing after intensive reading training?

Procedure: Auditory processing was investigated behaviourally and electrophysiologically (using MMN) in children 8-12 years. The behavioural battery involved tests such as the Frequency Pattern Test, Dichotic Digit Test, Random Gap Detection Test and Speech-in-noise. Electrophysiological tests involved MMN evoked by pure tones, chords and two speech stimuli. Reading fluency and accuracy, receptive vocabulary, and phonological awareness (nonword reading) were also assessed. The control group had 21 children with no reading problems. The group of children with reading disorder included 23 children with a reading delay of at least two years, recruited from a special education centre.

For the MMN recordings there was a repetitive “standard” stimulus, with a “deviant” sound present randomly in the sequence. MMN is the response to the change from the standard to the deviant sound. The MMN response to tonal stimuli was measured using a difference waveform created when the response to the standard was subtracted from the response to the deviant (Figure 1). For speech stimuli, the deviant stimulus was presented alone as well as interspersed with the standard stimulus. MMN was extracted by subtracting response to the deviant when presented alone (thus standard, no change of stimulus) from the response that was presented in oddball paradigm (therefore deviant, as the stimulus changed randomly). MMN responses were recorded using the NeuroScan evoked potential system via scalp electrodes placed at the vertex midline (Cz), frontal midline (Fz), and right (F4) and left (F3) hemispheres. During electrophysiological testing, the children watched a video of their choice with the volume turned down and were asked to ignore the stimuli that were being presented. All children’s peripheral hearing was assessed and found to be normal, using pure tone audiometry, immittance audiometry and transient click-evoked otoacoustic emissions.

Note: MMN can be elicited using two different extraction techniques. One is illustrated above and is often used with tonal stimuli. Another technique requires two different blocks of stimuli, one is an oddball (as shown being delivered to the subject) and the other block has only the deviant. Thus MMN would be elicited by subtracting the deviant alone from the deviant in oddball (often used with speech stimuli).

In a separate training study, a small group of children (N=6) with reading levels that were at least two years behind their chronological age were tested at the start and at the end of a school term. During the term, all six children received intensive reading intervention 5 days per week that involved working on phonological processing (phonics, blending, rhyming, and segmenting), reading and comprehension.

Results and Discussion: As expected the reading fluency and accuracy scores for the children with reading disorder were significantly poorer than the control group. The control and reading disorder groups differed significantly on most test results assessing reading fluency, accuracy, vocabulary, nonword reading, and behavioural test battery (Figure 2). For the behavioural APD test battery, more reading-disabled children had problems with the Frequency Pattern Test than with any other
The Frequency Pattern Test measures temporal sequencing as well as frequency discrimination. Thus further study would need to be undertaken to assess whether it is temporal sequencing or frequency discrimination that is affected in children with reading disorder. Jerger and Musiek (2000) suggested random gap detection as a clinical test of temporal processing for children with suspected APD. Although children with a reading disorder had significantly poorer gap detection results, the scores were still within normal limits. This was an interesting finding, since it has been suggested (e.g., Tallal et al., 1993) that children with a reading disorder have a specific temporal processing problem affecting their ability to discriminate rapidly presented sounds.

Mismatch negativity is a very small response (amplitude similar to ABR, Picton et al, 2000). MMN is not very robust and can be absent in normal populations (e.g., Dalebout and Fox, 2001, Wunderlich and Cone-Wesson, 2001). This may have been a factor contributing to the overall lack of group differences for MMN. Only the speech stimulus /ga/ produced significant differences in MMN area between the groups. Figure 3 and 4 depict area and waveforms for children in the two groups who elicited MMN for the tonal and speech (/ga/) stimulus. Like the behavioural FPT task, the positive results for the /ga/ stimulus could reflect the fact that the detection of the stimulus change from standard to deviant could be based on both temporal and spectral cues. All children with a reading disorder showed auditory deficits. Thus, it is recommended that all children with reading disorder be tested for APD, and similarly children with APDs should have their reading and phonological awareness abilities investigated.

In the reading training study, all six children showed evidence of APD based on both behavioural and electrophysiological measures prior to the intervention. The children then underwent intensive
training with emphasis on listening comprehension, phonological awareness, and cognitive strategies provided at a special education centre by reading specialists. There were significant improvements in reading fluency and accuracy after training. The behavioural APD test battery and MMN did not differ significantly pre- versus post- training. The obligatory responses (P1-N1-P2) showed significantly enhanced P2 amplitude after the training. There was no control group but the individual changes in P2 were compared to maturational norms that are available in the literature. Ponton et al’s (2000) data shows very little change in P1-N1-P2 latencies and amplitudes over 6 months in the 10-12 years of age group. Thus, the change in P2 amplitudes seen in four of the six individual children could not be explained on the basis of maturation alone. The changes observed for the auditory electrophysiological responses following listening training are not unprecedented. Other studies have also shown improvements in cortical response amplitudes after auditory training (Tremblay et al., 2001; Atienza et al., 2002; Hayes et al., 2003; Reinke et al., 2003).

Conclusions and Implications: This research has shown that children with reading disorders are likely to have a co-occurring auditory processing disorder. Consequently, all children with a reading disorder should have an auditory processing evaluation before designing intervention. The presence of co-existing auditory processing disorders would suggest that intervention should be modified so that the auditory processing deficits are also addressed in the remediation. Children participating in the training study had a coexisting APD and the reading training undertaken with the children had a major emphasis on listening comprehension. It is also recommended that children with auditory processing disorders have a detailed reading evaluation. The typical pattern of auditory processing deficits in children with a reading disorder in our studies mostly appears to be a problem with temporal sequencing and discrimination (e.g., Frequency Pattern Test). If electrophysiological data can also be collected then there may be an absent mismatch negativity to the speech sound /ga/. Future study will be undertaken to see if the presence of poor temporal pattern processing and absent /ga/-evoked MMN are reliable diagnostic features of APD in children with reading disorder. Lastly, an intense reading training program that encompasses auditory training (including listening comprehension), phonological awareness and metacognitive strategy training may result in changes in auditory processing. More research is needed to understand the nature of these changes.

Jessica smiling through the electrophysiological assessments

References


The Evaluation of Alternative Stimuli for Hearing Assessment of Young Children: A Computer-aided Assessment Program

Investigators: Robyn Massie, Harvey Dillon

This study is supported by CRC Hear.

Background: Clinical audiologists have long expressed a need for the development of stimuli alternative to the conventional pure tones and frequency modulated warble tones when assessing young children. The motivation for using other stimuli stems from the fact that young children have short attention spans which leave a limited time window for testing that gets shorter when uninteresting stimuli are used (Besing, Koehnke, Abouchacra, & Letowski, 1998). Visual Reinforcement Audiometry has been shown to be a reliable and efficient assessment procedure to use with children aged from five months to three years. As the term suggests, Visual Reinforcement Audiometry is an operant conditioning technique which capitalises on a child’s natural inclination to turn towards a sound source. Typically, a 90-degree head-turn response after a discriminative stimulus is rewarded with an interesting visual event, usually activation of a lighted animated hand puppet presented from behind a one way screen (Cichello, 1985; Haug, Baccaro, & Guilford, 1967: Widen, 1990).

Infants respond most to complex broadband signals, less to low frequency signals, and least of all to high frequency signals (Hoverstein & Moncur, 1969; Thompson, & Thompson, 1972). In 1985, Thompson and Folsom investigated auditory-stimulus effects on the head-turn response behaviour of infants under...
conditions where responses were either reinforced or not reinforced with visual stimulus. They found that bandwidth characteristics of the auditory stimulus had no influence on response behaviour after conditioning had been established.

A broadband stimulus commonly used for assessing the behavioural responses of infants and young children is speech. For example, in their Visual Reinforcement Audiometry protocols, Sabo, Paradise, Kurs-Lasky and Smith (2003) obtained speech recognition thresholds before presenting warble tones. Widen et al., (2000) also included broadband speech in their protocols for insert earphone Visual Reinforcement Audiometry, the reason given to corroborate their pure tone findings.

In an attempt to find natural and familiar sounds which may prove more interesting to young children, audiologists have employed other sounds such as noisemakers (e.g., bells, rattles) and environmental sounds (e.g., animal sounds, baby cries) in their assessments (Bove & Flugrath, 1973; Northern & Downs, 1984). In 1996, Myers, Letowski, Abouchacra, Kalb, and Haas set out to determine whether selected sound effects could be used as alternative stimuli to pure tones when testing children and other special populations. The authors found that thresholds obtained with some environmental sounds octave-band filtered at 250 Hz, 500 Hz, 1000 Hz, 2000 Hz, and 4000 Hz were within 2-3 dB of pure tone thresholds. They concluded that filtered sound effects could be a promising solution to pure tone stimuli for use in audiometric tests.

In 1989, Birtles developed a Visual Reinforcement Audiometry screening procedure aimed at identifying hearing-impaired or difficult-to-test children from the large numbers being referred to audiology clinics in New South Wales. Designed to be efficient, have low false positive rates and no false negative cases, the stimuli used in this procedure were 1000 Hz and 4000 Hz warble tones presented at screening levels of 30 dB SPL and 25 dB SPL respectively. A slightly modified version of this screening procedure was used in the present study.

**Objectives:** This study forms part of a larger project to develop a computer-based unit for the behavioural assessment of hearing in young children. The aim of this study was to examine whether speech sounds and noisemakers filtered at 1000 Hz and 4000 Hz were as effective, worse or better than conventional warble tones. The hypothesis was that normally hearing young children from six months of age will respond more frequently and/or show fewer false positive responses to speech or other types of stimuli than to tonal stimuli.

**Procedure:** Forty-two subjects between the ages of six months and thirty months were required for the study. Only children who were considered to be developmentally normal and had no known history of middle ear problems were considered for participation. A total of 65 children were assessed before the required number of 42 final subjects, without evidence of middle ear dysfunction in one or both ears, were achieved. The final forty-two subjects comprised 26 males and 16 females, and ranged in age from 10 months to 28 months (mean = 18 months).

Each audiological assessment included otoscopy, tympanometry, and Visual Reinforcement Audiometry. A wide range of stimuli comprising speech sounds and noisemaker sounds were recorded on compact disks. These stimuli were one-octave band filtered to include the dominant frequency region. The filtered stimuli were circulated to a group of experienced pediatric audiologists who were asked to choose the stimuli they felt would be most intrinsically interesting to infants. The stimulus set was reduced to a final six following audiologists’ judgements: filtered speech at 1000 Hz (/ala/) and 4000 Hz (/s/), filtered noisemakers at 1000 Hz (a synthetic high frequency stimulus) and 4000 Hz
(a non-reed squeaker), and the 1000 Hz and 4000 Hz warble tones. All stimuli were presented to the 42 infants, with two familiarisation trials at 65 dB SPL being given for each stimulus prior to it being presented three times at test levels.

The same two examiners performed the audiological assessments. The examiner in the control room was an experienced pediatric audiologist. The examiner in the test room was a tertiary-qualified research assistant who had extensive prior experience with young children. Each child sat on a parent’s lap or independently on a chair one meter from a loudspeaker located at a 90 degree angle from the child’s front line of vision. A visual reinforcer box was located above the loudspeaker on the other side of the one-way mirror. A variety of hand-held puppets acted as reinforcers. A response was defined as a head turn towards the loudspeaker/visual reinforcer box within four seconds from the onset of the stimulus. To account for any order effect, the six stimuli were assigned to subjects using a Latin square design.

**Findings:** This project is in progress and the following presents some initial findings. The results of statistical analysis are outlined in [Figure 1](#).

**Figure 1.** The number of times infants responded to each stimulus. Error bars show 95% confidence intervals.

The major outcome measure was the number of times infants responded to each stimulus. There was a significant stimulus effect, with the speech stimulus /s/ at 4000 Hz being significantly worse than the other five stimuli. There were no other significant differences. For the speech stimulus /s/, infants responded two out of three times on average, whereas the other four stimuli were responded to on average 2.6 times out of three. These results illustrate that both noisemaker stimuli (synthetic high frequency stimulus and a non-reed squeaker) and the speech stimulus /ala/ were equally effective to 1000 Hz and 4000 Hz warble tones, the speech stimulus, /s/ was less effective, and none were more effective.

**Significance:** The aim of this study was to examine alternative stimuli that could be used to perform the Visual Reinforcement Orientation Audiometry screening procedure. The majority of non-tonal stimuli were no more nor no less effective than conventional warble tones in eliciting responses. Perhaps it is not surprising that stimuli which have the same short-term rms level and the same bandwidth are nearly equally attractive to children. However the question must be asked as to why the speech stimulus /s/ was not as effective as the others, particularly in view of Thompson and Folsom’s (1985) findings that, once conditioning had been established, the bandwidth of the signal did not affect threshold. As noted by Birtles (personal communication 2003), one of the difficulties in judging the superiority of stimuli is that some sound natural when unfiltered but, in some filtered conditions, lose their intrinsic attractiveness or natural character. This may have been the case with the speech stimulus /s/. Alternatively, unlike the other five stimuli, its spectral shape did not vary systematically with time over the duration of the spectrum.

Although this project was not designed to investigate the middle ear status of a cohort of infants, mention should be made of the 35% of potential participants who failed the protocols due to middle ear dysfunction. The average hearing loss
for this cohort of children was 39 dB SPL, a mild hearing loss. This data provides some evidence of the incidence of unidentified middle ear problems in this Caucasian population, and supports reports in the literature that middle ear effusion is prevalent in this age range (Teele, Klein, & Rosner, 1989). It was recommended that these children seek reevaluation following medical advice to exclude the presence of a permanent hearing loss.

With the advent of universal screening, children are being diagnosed with hearing loss at an earlier age and therefore require accurate threshold assessment. A primary limitation of Visual Reinforcement Audiometry when used in the traditional manner is that it is personnel-intensive and expensive (Widen, 1990). The findings from this study will contribute to the incorporation of suitable non-tonal stimuli in the HEARLAB computer-based system presently being developed by the National Acoustic Laboratories. This audiometer will include an option whereby both Behavioural Observation Audiometry and Visual Reinforcement Audiometry can be conducted by a single clinician. Clinical audiologists have indicated a need for the development of a variety of alternative stimuli to assess infants and young children, particularly those who are difficult to assess due to intellectual and/or physical difficulties. The next phase of this project will investigate whether a diversity of stimuli at each frequency, rather than a single stimulus, assists in maintaining the interest of the child for a longer duration.

References


Assessing workplace “Safety Climates”

Investigator: W Williams and SC Purdy

Background: When carrying out an OHS or safety survey there are a lot of questions that one would like to ask. Due to restrictions of space, employees’ time and in particular the attention span of the respondents, questionnaires or interviews must be restricted in length. Operating under just such restraints the authors of the current work were required to reduce the size of a large ‘safety climate’ questionnaire while retaining reliability and validity. The result was a series of six items with excellent reliability and validity compared to the original, larger questionnaire. The development of such a short questionnaire allows those interested to examine an organisation’s safety culture without the need for a larger research tool.

Occupational health and safety workers have been aware for many years that there is a direct relation between employee perceptions of safety within an organization and the safety performance of the organization (Bailey and Peterson: 1989; Shaw and Blewett: 1996; Hale and Hovden: 1998). These perceptions of safety give rise to what has been referred to as the safety climate of an organization (Zohar: 1980). Efforts have been directed at validating, assessing and measuring the safety climate of organizations (IAEA: 1991; Cooper and Phillips: 1994; IAEA: 1996; Du Pont: 1997; Pitzer: 1999). One questionnaire (Williamson, Feyer, Cairns and Biancotti: 1997) developed and implemented by the National Occupational Health and Safety Commission, has been shown to be a particularly suitable tool. This Safety Climate questionnaire contains seventeen questions.

In many occupations, workplace injuries such as hearing loss, are ‘accepted’ as an inevitable consequence of the work (Milhinch and Dineen: 1997). The belief that accidents and illnesses are natural consequences of work has been referred to as “fatalism” (Pitzer: 1999). Fatalism is a barrier to achieving safety within the workplace since people with fatalistic attitudes will accept high injury and disease rates as inevitable and unavoidable (Shaw and Blewett: 1996). In the construction industry Milhinch and Dineen (1997) found that individual workers were more interested in protecting themselves from the immediate hazards of physical injury and were less concerned with hazards that may have future, uncertain negative outcomes.

Personal factors will influence the success of safety programs. Studies have demonstrated that the use of hearing protectors is significantly affected by perceived self efficacy, noise annoyance, perceived barriers to and benefits from hearing protector use, and perceived personal susceptibility to hearing loss (Melamed Rabinowitz, Feiner, Weisberg and Ribak: 1996; Kerr, Lusk and Ronis: 2002; Williams, Purdy Murray, Dillon, LePage, Challinor and Storey: 2004). Self-efficacy refers to a person’s belief in their capability to successfully perform a particular task (Bandure: 1977). For example, self-efficacy is an important predictor of hearing protector use (Lusk, Ronis and Hogan: 1997) and is also likely to determine whether people use other means to reduce their noise exposure such as engineering or administrative controls.

There is increasing recognition that safety solutions solely based on engineering approaches and legislation will fail if attitudes to safety are poor and effective safety management systems are not in place (Williamson, Feyer, Cairnes and Biancotti: 1997). The term "safety culture" or "safety climate" is used to describe the shared attitudes and perceptions employees have of the work environment that influence the behavior of people within the organization (Williamson, Feyer, Cairnes and Biancotti: 1997; Pitzer: 1999). There is a direct relationship between OHS management and the number of compensation claims (Gallagher: 1994) thus the integration of occupational health...
and safety into management systems is essential for improving OHS performance in industry (Gallagher: 1997). The relationship between safety climate and accident rates has been investigated (Bailey and Paterson: 1989) but to our knowledge no study has investigated whether safety climate influences the effectiveness of hearing loss prevention programs.

**Research aims:**

The research aims of this project were:-

a) to assess workplace safety climate using an existing assessment tool (Williamson, Feyer, Cairnes and Biancotti: 1997);

b) to compare the measured safety climate to an assessment of the perceptions to noise exposure in the workplace using the “Noise at Work” questionnaire (Purdy and Williams: 2002); and

c) to develop a shortened version of the safety climate assessment tool while maintaining validity and reliability.

**Procedure:** Subjects were 69 workers (43 men, 25 women) from two separate workplaces, A and B. The average age was 41.4 years (SD 12.1). Workplace A had 13 in the training group and 17 control subjects. For Workplace B there were 22 in the training group and 17 control subjects. Seventy-four percent of all subjects spoke English at home. Overall work experience was 8.6 years (SD 8.5) and the average duration in the current job was 7.4 years (SD 8.3).

Workplace A utilises rotating shifts of work groups of five to ten people and these groups are spread over several different work locations. The noise is restricted to specific areas and is in the range of 95-100 dB(A). Workplace A has a well-established, comprehensive OHS program with regular training provided by the organisation. Individuals are well trained for their specific and general tasks within their work area and operate as teams with a team leader rather than in a usual supervisor – worker relationship.

Workplace B contains around 100 employees all on one site which consists of a large single building. Some areas of the building are clearly divided from the main work area, for example in the office spaces and executive area. However, in the main, noisy machinery is distributed throughout the work area, resulting in noise levels that range between 57-85 dB (A) [average 76 dB]. Staff at Workplace B have minimal OHS skill levels and some informal, job-specific training.

The ‘safety climate’ questionnaire is intended to give a measure of the safety climate of a workplace through prevailing attitudes and perceptions. It is not directed to any specific workplace hazard but overviews workplace safety in general. This is done through 17 items (Williamson, Feyer, Cairnes and Biancotti: 1997) which consider five major factors: Personal motivation for safety; Positive safety practice; Risk justification; Fatalism; and Optimism. The questionnaire has a five-point Likert scale for most items (e.g. “I cannot avoid taking risks in my job”; and “Accidents will happen no matter what I do”) along with several open questions.

Along with the Safety Climate questionnaire the Noise at Work questionnaire (Purdy and Williams: 2002) was also used to assess the specific workplace hazard of noise exposure. This questionnaire has five sub-scales that assess:-

1. **Benefits** – perceived benefits of reducing noise and noise exposure, for example, using hearing protectors or other methods to reduce noise exposure;

2. **Barriers** – perceived barriers to reducing noise exposure;

3. **Self-efficacy** – perceived personal ability to reduce noise exposure and/or protect hearing (Bandura: 1986);

4. **Attitude** – attitudes to workplace noise and noise exposure; and

5. **Susceptibility** – an individual’s perceived susceptibility to hearing loss, ie, whether they think noise exposure can or will damage their hearing.
Demographic information was also collected and included: age, gender, level of education; type and duration of work experience. Information was also gathered on self-reported and family-reported hearing loss; experience of tinnitus; and noise exposure, use of hearing protectors and other methods of noise exposure avoidance at work.

**Findings:** Demographic and questionnaire responses were analysed using Mann-Whitney U and t-tests to examine differences between the two workplaces. Subjects from the two workplaces did not statistically differ in their age, language spoken at home, self-reported hearing loss and tinnitus, or self-rated level use of hearing protectors.

<table>
<thead>
<tr>
<th></th>
<th>Workplace A</th>
<th>Workplace B</th>
</tr>
</thead>
<tbody>
<tr>
<td>% male</td>
<td>97%</td>
<td>36%</td>
</tr>
<tr>
<td>Education level</td>
<td>73% above high school level</td>
<td>87% high school</td>
</tr>
<tr>
<td>Type of work</td>
<td>97% plant &amp; machine operators</td>
<td>18% plant &amp; machine operators</td>
</tr>
<tr>
<td></td>
<td>56% labourers</td>
<td></td>
</tr>
<tr>
<td>Previous hearing test</td>
<td>100%</td>
<td>38%</td>
</tr>
<tr>
<td>Job experience</td>
<td>12 years (std dev 10.5)</td>
<td>4 years (std dev 4.2)</td>
</tr>
<tr>
<td>Hearing protector use</td>
<td>18% (std dev 21.5)</td>
<td>8% (std dev 21.2)</td>
</tr>
<tr>
<td>Self-rated noise exposure</td>
<td>20% (std dev 21.8)</td>
<td>39% (std dev 36.9)</td>
</tr>
</tbody>
</table>

**Table 1.** Summary of workplace demographics for items that differed significantly between the two workplaces.

As shown in **Table 1**, there were significant differences between workplaces for gender, education level, type and level of work, job length, and previous hearing testing (p < 0.004). There were also significant differences between the two workplaces in self-rated amount of noise exposure (p = 0.007), safety scores (p < 0.001), and in their Noise at Work questionnaire scores for all five sub-scales (p ≤ 0.045).

The differences in scores between workplaces for both the Noise at Work and Safety Climate questionnaires are shown in **Figure 1**. The Benefits, Barriers, Self-efficacy, Attitudes and Susceptibility sub-scores arise from the “Noise at Work” questionnaire while the Safety score comes from the Safety Climate questionnaire.

The original Safety Climate questionnaire contained 17 items to which participants assigned a rating using a 5-point “strongly agree” to “strongly disagree” scale for 12 items and a 5-point “always” to “never” scale for the remaining five items. A reliability analysis (Cronbach’s alpha) was performed on all 17 items for the 66 subjects that completed the Safety Questionnaire. The alpha value for the 17 items was 0.69, which is comparable to a previously reported value of 0.61 (Williamson, Feyer, Cairnes and Biancotti: 1997).

By removing items with poorer item-total correlation, a subset of six items with very good overall reliability (α = 0.78) was obtained (**Table 2**). These six items all require a response on a 5-point scale, from 1 = strongly agree to 5 = strongly disagree. The mean rating for these 6 items was 3.4 (SD = 0.82). All six items gave a range of responses from 1 to 5 on the scale,
indicating a range of safety perceptions. Participants had average responses for each item between 3 and 4 indicating that they tended to disagree with the statements. The six items cover: recognition of workplace dangers (item 2); perception of safety as a priority (items 4 and 5); and fatalism about accidents/injuries (items 7, 8, 10). Higher ratings (disagreeing with statements) reflect better safety perceptions (ie, aware of dangers in the workplace, believe that safety is a priority all the time, believe that accidents can be avoided rather than being fatalistic). Out of the six items most people disagreed with items two and five, indicating that they were generally aware of dangers and thought that safety should be a priority. For items 7, 8, and 10 only about half disagreed with the statements, indicating that many participants were fatalistic about accidents. The Safety Climate questionnaire was demonstrated to be capable of reduction from 17 to six basic items while maintaining an excellent overall internal reliability rating (α = 0.78). Unfortunately due to time and workplace pressures there was insufficient opportunity to retest validity and test-retest reliability of the reduced questionnaire using respondents from the same workplaces.

The reduced number of questions in the Safety Climate questionnaire result in a shorter and more easily applied version of the Safety Climate questionnaire in workplaces where the individuals’ participation time is limited.

### References


<table>
<thead>
<tr>
<th>Item #</th>
<th>Item Wording</th>
<th>% agree</th>
<th>% disagree</th>
<th>Mean (σ)</th>
<th>Item-total correlation</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>In the normal course of my job, I do not encounter any dangerous situations.</td>
<td>23</td>
<td>72</td>
<td>3.62 (1.15)</td>
<td>0.44</td>
</tr>
<tr>
<td>4</td>
<td>Safety works until we are busy, then other things take priority.</td>
<td>32</td>
<td>59</td>
<td>3.51 (1.33)</td>
<td>0.57</td>
</tr>
<tr>
<td>5</td>
<td>If I worried about safety all the time, I would not get my job done.</td>
<td>19</td>
<td>69</td>
<td>3.75 (1.17)</td>
<td>0.63</td>
</tr>
<tr>
<td>7</td>
<td>I cannot avoid taking risks in my job.</td>
<td>34</td>
<td>53</td>
<td>3.25 (1.24)</td>
<td>0.57</td>
</tr>
<tr>
<td>8</td>
<td>Accidents will happen no matter what I do.</td>
<td>34</td>
<td>49</td>
<td>3.19 (1.25)</td>
<td>0.48</td>
</tr>
<tr>
<td>10</td>
<td>Not all accidents are preventable, some people are just unlucky.</td>
<td>28</td>
<td>48</td>
<td>3.18 (1.07)</td>
<td>0.48</td>
</tr>
</tbody>
</table>

Table 2. Wording of the six 'Safety' questionnaire items used to derive a 'Safety' score for the analysis of variance. Cronbach’s alpha for these six items was 0.78.
Instruction and the improvement of hearing protector performance

**Investigator:** Warwick Williams

**Background:** Too often, in spite of the encouragement of those who advocate the reduction of noise levels in the workplace as the preferred solution to noise exposure, hearing protectors are provided as the first and only line of defence against noise exposure. Unfortunately hearing protectors are too often supplied with no real instruction. This degrades their expected performance.


Training is more important when the prevention of noise exposure relies almost solely on the provision and wearing of hearing protectors. “[B]efore personal hearing protectors are issued, the need for their use should be fully explained” (NOHSC: 2000, Section 9.13). As discussed by Berger (2000) if subjects are not trained in the fitting of hearing protectors we can never be sure of their actual performance. In fact we can never be sure of the real performance of a hearing protector in practice unless measurements are made using a microphone placed appropriately in the ear. In practice the assumption tends to be made that when hearing protectors are being worn they are working at or around their rated capacity.
Many hearing protector acoustic attenuation performance measures, including the Noise Reduction Rating (NRR) (ANSI S 12.6 – 1997) and the Sound Level Conversion (SLC80) (AS/NZS 1270: 2002), utilise a figure that is a direct function of the mean attenuation and standard deviation measured for several Octave Bands. In the case of NRR this figure is a function of the mean attenuation minus two standard deviations while in the case of the SLC80 it is a function of the mean minus one standard deviation. Thus to achieve an NRR or SLC80 value that most accurately represents the performance of the device it is particularly advantageous to have a small standard deviation.

The descriptor used for comparing the performance of the attenuation of hearing protectors in Australia is the SLC80, expressed in dB. This descriptor, the Sound Level Conversion, is deemed to offer the specified amount of attenuation or greater to approximately 80% of the wearers at any one time. The SLC80 is a direct function of combination of the mean attenuation minus the standard deviation at the seven octave bands tested. The detailed method of calculation is given in AS/NZS 1270: 2002, Appendix A. the process is described more fully in Waugh (1984).

For a hearing protector to have an overall high SLC80 rating it should ideally have a high mean attenuation and low standard deviation. The worst case is a low mean attenuation with a corresponding wide standard deviation. For a well behaved device that offers high or low attenuation, the standard deviation at all octave bands should be as small as possible.

In this project the attenuation performance of one particular model of earplug was compared with and without instructions to test subjects.

**Research question:** Can a modest amount of instruction improve hearing protector performance?

**Procedure:** Measurement of the attenuation of the selected ear plugs was carried out using the NATA Certified facilities at NAL. The procedure followed was as detailed in Australian Standard AS 1270: 19881.

Subject recruitment and instruction required the use of 'inexperienced' test subjects, similar to ANSI S12.6 – 1997 Method B: Subject fit, draft ISO 4869-7 and the current combined Australian/New Zealand Standard AS/NZS 1270: 2002 Acoustics – Hearing protectors.

The distinguishing feature of this form of testing, “the subject-fit method”, is that the experimenter (tester) is not permitted to directly instruct the subject(s) on how to fit the device (ear plug, earmuff, etc). Any form of one-to-one instruction is excluded, except in the case where it is inherently built in to the supply of the device, for example in the case of personally moulded ear plugs. Instructions may only come from the information typically supplied to the consumer by the manufacturer/distributor. This may be in the form of a written and/or pictographic presentation.

Two series of attenuation tests were carried out on the one particular ear plug which was supplied in a single size only. The first series of tests involved fifteen test subjects with no fitting instructions supplied in any form. The method of fit was left up to the test subject. The test method does not allow any assistance from the tester to fit the device only the use of the supplied instructions. Fitting noise was supplied to allow the subjects to adjust the plug fit if they wished.

The second series of tests involved sixteen test subjects. This time instructions were supplied by the manufacturer/distributor. These instructions are illustrated in Figure 1 and are typical of those provided on the back of the small plastic ‘pocket’ in which many companies supply ear plugs.
Figure 1. An example of the simple word and pictographic information used by the second group of sixteen test subjects (the quality represents that typically found on the container)

**Results:** The overall test results for the two series of tests without and with manufacturer’s/distributor’s supplied instructions are provided in Tables 1 and 2 respectively. As can be seen the results with no instructions have much greater variation resulting in greater variance and an overall lower SLC\(_{80}\) rating, 8 dB compared to 24 dB as would be more usually expected for this type of plug.

### Frequency (Hz)

<table>
<thead>
<tr>
<th>Subject</th>
<th>125</th>
<th>250</th>
<th>500</th>
<th>1k</th>
<th>2k</th>
<th>4k</th>
<th>8k</th>
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<tbody>
<tr>
<td>1</td>
<td>2.1</td>
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<td>2</td>
<td>17.8</td>
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<td>14.2</td>
<td>19.5</td>
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<td>38.2</td>
<td>33.4</td>
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<td>3</td>
<td>14.5</td>
<td>11.5</td>
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</tr>
<tr>
<td>Mean – SD</td>
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<td>4.2</td>
<td>10.8</td>
<td>9.5</td>
<td>12.2</td>
</tr>
</tbody>
</table>

**Table 1.** Octave band test results for earplug with no fitting instructions provided.

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![Figure 2](image_url)

**Figure 2.** Number of test subjects versus attenuation at 4kHz for “instructions supplied” and “instructions not supplied”
The overall performance of the plugs increased significantly with the provision of very simple instructions. For all frequencies, all of the means increased and the standard deviations decreased resulting in an overall increase in SLC<sub>80</sub> rating of 16 dB, from 8 to 24 dB.

Not all subjects experienced attenuation performances at the same level. For example, in Table 2, subjects 5 and 11 provided attenuation figures well below the mean, while subjects 4 and 16 provided values, in general, well above the mean. Thus while some or better instruction may help decrease the probability of a "bimodal distribution" of results (Williams: 2003) it may not necessarily provide the complete solution. The extrapolation that if a little information is good therefore more will be better may not necessarily be the case. Providing more detail and sometimes complexity has been shown not to increase preventative action in noise exposed individuals (Williams, Purdy, Murray, Dillon, LePage, Challinor and Storey: 2004). Any instruction must be tailored to the situation and the individuals’ needs.

**Significance:** If hearing protectors are to be utilised as the first line of defence against noise exposure then wearers should be instructed and educated in their use in order to ensure rated and consistent performance is experience. With the very simple information provided in this case the increase in performance of the earplugs tested was significant.

### References


1 Note: This Standard has undergone subsequent changes, the main one being that testing is now required to use pink noise of third-octave band width, at octave band centre frequencies. Previous testing was carried out utilising a pure tone. “For practical purposes pure-tone and 1/3-octave band measurements of ear-protector attenuation are identical.” (Waugh: 1974, p 1868)
**Hearing Loss and Tinnitus Simulation Compact Disc**

**Investigators:** Michael Fisher, Nicky Chong-White and Harvey Dillon

Conveying the effect of hearing loss to someone with normal hearing can be a challenging task for practitioners in the field of hearing conservation and rehabilitation. To assist practitioners, NAL has produced a compact disc (CD) that contains simulations of the effect of tinnitus and loss of hearing sensitivity. The CD may be used to demonstrate to the friends and relatives of a hearing-impaired person the effect that hearing loss has on their ability to perceive sound, and the internally generated sounds that they may hear. The CD may also be used in a hearing conservation program to demonstrate to normal-hearing individuals the effect that excessive noise exposure can have on their hearing, or the hearing of others.

The CD contains a range of stimuli suitable for both objective and subjective demonstration of hearing loss. The stimuli include two AB word lists, two BKB sentence lists, discourse in both quiet and noisy backgrounds, and samples of contemporary music. The stimuli may be used with normal hearers to test their ability to understand speech that has been changed to simulate hearing loss, thereby demonstrating the reduction in objective performance caused by hearing loss.

Each set of stimuli has been processed to simulate the effect of four different degrees of hearing loss. The stimuli in their unmodified form are also present on the CD for comparative purposes. It is intended that the unmodified reference stimuli be presented at an average sound level of 65 dB SPL. With this reference sound level in mind, the hearing loss types have been chosen so that the stimuli remain audible to some degree in a quiet environment to normal hearers, while conveying various degrees of perceptual difficulty that come with hearing loss. The four hearing losses simulated are: a flat 40 dB loss, a flat 25 dB loss, a 30 dB/octave sloping loss beginning at 1 kHz and a 30 dB/octave sloping loss beginning a 2 kHz as displayed in **Figure 1**.

The hearing loss simulation processing employs...
multi-band, fast-acting, dynamic-range expansion to simulate loudness recruitment. The expansion is applied independently within the separate frequency bands. The bands have bandwidths conforming to the critical band (bark) scale of hearing. At some high sound level hearing-impaired individuals (except those with a profound loss) will report the same sensation of loudness as normal-hearing individuals. In these simulations, this high sound level is set to 100 dB SPL although none of the stimuli reach this level. Figure 2 shows the relationship between the input and output sound levels at 2 kHz for 25 dB and 40 dB hearing loss simulations. Each band has an individual input-to-output relationship determined by the hearing loss within that frequency region.

No direct attempt has been made to simulate the temporal or frequency smearing aspects of hearing impairment. However, the frequency-specific dynamic-range expansion processing on the CD inherently increases both temporal and frequency masking by expanding the contrast in both the temporal dynamics and level differences across frequency.

The CD contains simulations of a range of tinnitus type sounds for demonstrating the distressing and debilitating effect of tinnitus. These include high frequency sounds such as high-frequency, narrow-band noise, similar to the sound made by cicadas and often associated with cochlear damage. Also included are lower frequency sounds such as a low frequency pulsing noise which is often associated with abnormal (muscle) activity in the middle-ear and blood circulation.

The CD may be used with any sound reproduction equipment which has a reasonably flat frequency response. A sound level meter is required for absolute sound level calibration of playback system. The CD contains a speech shaped noise signal for the purpose of calibration. Also included with the CD is a short booklet containing an example of a demonstration/training session.

Copies of the CD may be ordered via the NAL website: www.nal.gov.au, or from the Research Administrative Officer
Phone: +61 2 9412 6872
Fax: +61 2 9411 8273
E-mail: research@nal.gov.au

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* With permission from La Fiesta Sound System (© 2004 L. Foster/J Shave/H. Walker)

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Current Limitations of Hearing Aid Batteries

Investigators: Eric Burwood, Walter Crichton

Background: The National Acoustic Laboratories has tested hearing aid batteries for over 30 years. During that time there have been significant advances in battery technology resulting in better performance and this is particularly so for button cells used in behind-the-ear (BTE) and in-the-ear (ITE) hearing aids. When first introduced, the zinc-air technology doubled the battery capacity of the battery types used at that time and performance is still improving in various ways.

Research Questions:
1) What electrical current is required for contemporary hearing aid and cochlear implant technology?
2) What electrical current can commercially available hearing aid batteries supply?
3) What happens to a hearing aid if the battery voltage momentarily drops?
4) Do existing hearing aid batteries meet the requirements of hearing aids?
5) Are existing battery testing methods adequate?

Findings:
1) For high-powered hearing aids, the electrical current fluctuates as the input sound level varies. For hearing aids using a size 675 zinc-air battery, maximum current levels between 20 and 25 milli-ampere have been measured when the input sound pressure level (SPL) was between 70 and 90 dB SPL. For hearing aids using a size 13 battery, maximum currents in the vicinity of 10 to 12 milli-ampere have been measured. In the samples tested, the high current occurred in particular frequency zones when the hearing aid was adjusted to high power and volume control settings. With no acoustic input the current level drops considerably to the quiescent level that may be of the order of one or two milli-ampere.

For cochlear implants the current requirement can vary depending upon how the device is set up, however the current levels may be relatively constant and at a level of up to 20 milli-ampere.

2) Many hearing aid batteries have difficulty supplying high levels of electrical current for any significant length of time. Under a constant electrical load, the performance of different brands and models of zinc-air batteries varies considerably. For example, some size-675 batteries may be able to supply continuous currents up to 14 milli-ampere while others can supply 20 milli-ampere. Some size-13 zinc-air batteries may be able to supply continuous currents up to 7 milli-ampere while others can supply 10 milli-ampere.

3) For many hearing aids the battery voltage fluctuates as the input signal varies. This effect is more significant in medium and high power hearing aids. Consider a hearing aid set for microphone input. When a loud sound occurs the hearing aid electronics demands more power from the battery. If this situation persists for a length of time then the battery voltage will fall during that time. If the voltage falls below a reset voltage level, which is dependent upon the hearing aid electronic technology used, then the function of the hearing aid can be disturbed. When the loud input sound level returns to a more normal level then the demand on the battery is reduced and the battery voltage increases. As the increasing battery voltage passes through the reset voltage level, the programmable or digital hearing aid is reset to the initial settings determined by the fitting audiologist. If the hearing aid user had previously changed a control, such as the volume control, then the hearing aid user would perceive a different gain setting as the hearing aid is reset to initial power-on conditions. To the hearing aid user this change in volume setting may be perceived as an unwanted change and may be perceived as an intermittent fault with the hearing aid. If this hearing aid was brought in for repair, then no fault would be found. In this case, the reason for
this perceived fault would be that the battery could not supply sufficient current required by the hearing aid and hold the supply voltage above the reset voltage for the hearing aid. The reset voltage for the hearing aid is below the battery end-point voltage, generally taken to be of the order of 1.1 Volt.

4) For some hearing aid users who require a low to moderate acoustic output power, many of the currently available batteries will provide adequate performance. However, for those hearing aid users who require higher output levels, then only the better performing batteries will provide reliable operation under all listening conditions. Similarly for cochlear implants, where a high level of continuous electrical current is required, then only the better high-performing batteries will provide reliable operation.
If a lower performing battery is used in a device that requires high current, then unreliable operation may occur. For hearing aid users, this unreliable operation will occur under high continuous noise environments, such as in noisy industries, subway trains, or, near roads where there is a continuous high level of noise, particularly where heavy vehicles are operating. Also some leisure pursuits have high continuous noise levels.

5) Many years ago hearing aid batteries were tested under continuous light levels of load that were adequate for the available hearing devices at that time. Then as current demands increased, testing was modified to partly simulate the “in use” requirements by loading a battery with a continuous light load and periodically increasing the load to reflect the real life demands. This worked well for analogue hearing devices of the 1980’s and provided an indication of the total energy that could be usefully extracted from the battery. However with the introduction of programmable devices in the early 1990’s and digital devices in the late 1990’s, these existing testing methods did not reflect the real life demands, especially in those situations where sound levels could be elevated for more than the 0.1 seconds duration of the high current test load. This led the National Acoustic Laboratories to more stringent testing requirements in which the end of battery life occurs when hearing aid operation becomes unreliable, rather than when all useful energy is drained from the battery.

**General Observations:** Since the introduction of hearing aid zinc-air batteries, it has been noted that under similar testing conditions, the life for each size of zinc-air battery type has increased. Leakage and salting residues from these batteries have been less of a problem in recent years than initially observed when this battery technology was first used in hearing aids.

**Significance:** The significance of this work is that an improvement in hearing aid technology from analogue, to programmable, to digital hearing aids has created the need to re-examine the test methods for zinc-air hearing aid batteries. The battery testing procedures currently used by the National Acoustic Laboratories are designed to guarantee reliable operation of the hearing device and to allow the ranking of batteries under stringent test conditions.

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**Signal processing in modern hearing aids and localisation performance**

**Investigators:** Gitte Keidser, Kristin Rohrseitz\(^1\), Harvey Dillon, Volkmar Hamacher\(^1\), Uwe Rass\(^1\), Lyndal Carter, and Elizabeth Convery.

\(^1\) Siemens Hearing Instruments, Germany

*This study is supported by Siemens Hearing Instruments.*

**Background:** The cues for localising sounds in the horizontal plane include interaural difference in
time and level that are mainly used to discriminate between sound sources located to the left or right of the median plane and the spectral shape above 4 kHz that is mainly used to discriminate between sound sources located to the front or rear of the frontal plane. Studies on the effect of hearing loss on localisation performance have demonstrated that as long as the function of the cochlear is not too distorted and the presentation level is adequate, localisation performance in the horizontal plane is close to normal (e.g. Hausler et al., 1983; Noble et al., 1994). On the other hand, several studies have shown that localisation performance by hearing-impaired listeners aided is somewhat deteriorated relative to their unaided performance (e.g. Orton & Preves, 1979; Byrne et al., 1992; Köbler & Rosenthal, 2002). This is thought to be because time delay in hearing aids due to tubing, transducers, and processing can distort the interaural time difference cue (ITD) and because inadequate amplification above 4 kHz and the microphone position on behind-the-ear (BTE) devices can distort spectral cues. Most of the past research on aided localisation performance is based on analogue hearing aids and linear amplification whereas today’s digital hearing aids offer far more sophisticated signal processing that have the potential for further distorting the horizontal localisation cues.

Objectives: The aim of this study was to investigate the effect of multi-channel wide dynamic range compression (WDRC), noise reduction (NR), and microphone directionality on the horizontal localisation ability of bilaterally fitted hearing-impaired subjects.

Procedure: Sixteen normal-hearing and twelve hearing-impaired subjects with sensorineural loss that in terms of the pure tone average (PTA) ranged from 32 to 66 dB HL participated in the study. The hearing-impaired subjects were all fitted with a pair of Triano S BTE devices according to the NAL-NL1 prescription. Medium noise reduction and adaptive directionality were activated for daily use. Horizontal localisation performance was tested two weeks and two months post fitting.

The instrumentation for localisation testing consisted of twenty speakers arrayed 18 degrees apart in a 360 degree horizontal arc. A screen of black, acoustically transparent fabric enclosed the speaker array so that the speakers could not be seen from inside or outside the array. An additional speaker was mounted at 80 degrees for the presentation of a constant speech-shaped random noise, referred to as an activation noise. Stimuli presentations were software controlled, including randomisation of the presentations through twenty speakers, randomisation of a ±3 dB roving effect applied to the intensity level, and number of repetitions. Two test stimuli were used, one being a train of four pulses of pink noise with a 150 msec pulse duration and a 50 msec interpulse interval, and the other a 2 sec speech sample.

Nine test conditions were implemented and are listed in Table 1. In three of the conditions the hearing-impaired subjects wore linear amplification, the NR was switched off and the microphones were in omni mode. In these three conditions, the subjects were tested with pulsed pink noise presented at 65 dB SPL, the speech sample presented at 65 dB SPL, and the pulsed pink noise presented at 72 dB SPL with the activation noise on at 65 dB SPL, respectively. These conditions served at reference conditions for the three signal processing schemes. The normal-hearing subjects were tested unaided with the same three combinations of stimuli and presentation levels. Syllabic WDRC (no NR and microphones in omni mode) was tested with both pulsed pink noise and speech presented at 65 dB SPL. Noise reduction was tested on maximum (linear amplification and microphones in omni mode) using the pulsed pink noise at 72 dB SPL and the activation noise on at 65 dB SPL to ensure that the NR algorithm was activated throughout testing. Finally, three combinations of microphone modes across ears, including cardioid
mode on both ears, cardioid mode combined with fig8 mode, and cardioid mode combined with omni mode, were introduced to simulate some of the more extreme outcomes of using adaptive directionality. The different microphone mode combinations were tested only with pulsed pink noise presented at 65 dB SPL. For half the subjects the cardioid mode in the microphone mode mismatch conditions was fitted to the left ear and for the other half the cardioid mode was fitted to the right ear.

Table 1: An overview of the conditions tested with the hearing-impaired subjects.

<table>
<thead>
<tr>
<th>Amplification</th>
<th>Noise reduction</th>
<th>Microphones</th>
<th>Stimulus type/presentation level</th>
<th>Activation noise/presentation level</th>
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</thead>
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<tr>
<td>Linear</td>
<td>Off</td>
<td>Omni dir</td>
<td>Pink/65 dB SPL Off</td>
<td>Off</td>
</tr>
<tr>
<td>Linear</td>
<td>Off</td>
<td>Omni dir</td>
<td>Speech/65 dB Off</td>
<td>Off</td>
</tr>
<tr>
<td>Syllabic WDRC</td>
<td>Off</td>
<td>Omni dir</td>
<td>Pink/65 dB SPL Off</td>
<td>Off</td>
</tr>
<tr>
<td>Syllabic WDRC</td>
<td>Off</td>
<td>Omni dir</td>
<td>Speech/65 dB Off</td>
<td>Off</td>
</tr>
<tr>
<td>Linear</td>
<td>On (max)</td>
<td>Omni dir</td>
<td>Pink/72 dB SPL On/65 dB SPL</td>
<td>Off</td>
</tr>
<tr>
<td>Linear</td>
<td>Off</td>
<td>Cardioid</td>
<td>Pink/65 dB SPL Off</td>
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</tr>
<tr>
<td>Linear</td>
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<tr>
<td>Linear</td>
<td>Off</td>
<td>Omni/Cardioid</td>
<td>Pink/65 dB SPL Off</td>
<td>Off</td>
</tr>
</tbody>
</table>

During localisation testing, subjects were seated in a height-adjustable chair in the centre of the localisation array with the ears aligned with the speakers. The subjects were fitted with a headlamp and they were instructed to keep the beam from the lamp at the center of a zero degree marker situated straight in front of them whenever the sound was present. Head movements were allowed between sound presentations. Via a video monitor, the tester could see the position of the light beam and the subject’s head position in relation to an overhead marker. Subjects were instructed to verbally report (via an intercom) the perceived direction of the stimuli in degrees that was then keyed in to a PC by the experimenter. To assist the subjects, the directions were labelled in 10-degree intervals on the inside of the black screen covering the speaker array as well as on a map of the array. The direction straight in front of the subject was marked 0 degrees.

Findings: For each subject, visit and test condition, the total root-mean-square (RMS) localisation error was calculated from two responses to twenty

![Figure 1](image1.png)

**Figure 1:** The average total and modified RMS localisation errors measured across three test conditions for 16 normal-hearing listeners tested unaided and 12 hearing-impaired listeners tested aided with linear amplification 2 weeks and 2 months post fitting.

![Figure 2](image2.png)

**Figure 2:** The response pattern produced across three test conditions by a) 16 normal-hearing listeners tested unaided, and b) 12 hearing-impaired listeners tested aided with linear amplification. The broken line shows perfect localisation performance and the size of the bubbles indicates the number of observations behind each response.
directions. A modified RMS error was also calculated that ignored any front/rear confusions by projecting the presentation and response azimuths at the rear of the frontal plane to the mirrored azimuths at the front of the frontal plane. Figure 1 shows the average total and modified RMS errors produced across the three reference conditions by the normal-hearing listeners tested unaided and the hearing-impaired listeners tested with linear amplification two weeks and two months post fitting. Based on both the total and the modified RMS errors the hearing-impaired listeners with linear amplification produced significantly more localisation errors than did the normal-hearing listeners unaided ($p < 0.00002$). Relative to the average total RMS errors, the average modified errors produced by the hearing-impaired listeners were drastically reduced, suggesting that the hearing-impaired listeners tested with linear amplification generally made many front/rear confusions. This is presumably due to a combination of the microphone position on the test device and inadequate amplification at the higher frequencies. Figure 2 shows the response patterns produced by the normal-hearing and hearing-impaired subjects across the reference conditions. Whereas the localisation performance by the normal-hearing subjects generally is quite accurate, both front-to-rear and rear-to-front confusions are consistently made by the hearing-impaired subjects.

Figure 3 shows the average RMS error produced by the hearing-impaired listeners two weeks and two months post fitting with each of the three signal processing together with the relevant reference condition(s). With respect to multi-channel WDRC (Figure 3a) an ANOVA of repeated measures showed a significant effect of signal processing ($p = 0.04$). On average, the subjects made 2.8 degrees more localisation errors when tested with WDRC than when tested with linear amplification. Relative to the total RMS errors produced, this difference in performance was rather small. Further, the difference in performance was more pronounced two weeks than two months post fitting. Thus, it is questionable whether there is a real long-term effect of multi-channel WDRC on the hearing-aid user’s ability to localise sounds in the horizontal plane.

A similar analysis on the data related to NR (Figure 3b) also revealed a significant effect of signal
processing (p = 0.04). On average, the subjects made 3.3 degrees more localisation errors when tested with NR off than when tested with NR on. As before, this difference in performance was rather small relative to the total RMS error produced and the difference in performance was again more pronounced two weeks than two months post fitting. Therefore, it is also questionable whether there is a real long-term effect of using NR on the hearing-aid user's ability to localise sounds in the horizontal plane.

Finally, for directionality (Figure 3c), the statistical analysis revealed a significant effect of signal processing (p = 0.006) and a significant interaction between signal processing and acclimatisation (p = 0.04). Post hoc analyses showed that over time, the subjects significantly improved their localisation performance by 14 degrees, on average, when tested with a cardioid mode on both ears. Two months post fitting, the localisation performance was significantly better when tested with the cardioid pair and cardioid/omni combination than when tested with the omni pair and cardioid/fig8 combination. Further analysis of the data showed that front/rear confusions were reduced with the cardioid pair and cardioid/omni combination. This is presumably because the cardioid characteristic provides relatively less gain to the rear than to the front thus restoring some of the cues normally produced from the effect of pinna shading.

Equivalent analyses of the modified RMS errors (front/rear confusions ignored) showed that the subjects significantly increased the angle error when tested with WDRC relative to linear amplification (1.9 degrees on average), and when fitted with a microphone mode mismatch relative to when they were fitted with the same microphone mode on the two ears (9.8 degrees on average). It seems that when wearing a microphone mode mismatch, the subjects perceived the sound source to be nearer the ear wearing the microphone mode producing the least phase delay. Further, the angle errors produced with the cardioid/fig8 combination was significantly increased over time (4.1 degrees on average).

Overall, directionality had the most impact on the ability to localise sounds in the horizontal plane and over time, whereas the long-term impact of multi-channel WDRC and noise reduction was considered unimportant. This is presumably because directionality is the only signal processing of the three investigated that distort the ITD, which is believed to be the dominant cue for horizontal localisation for broadband sounds (Sandez et al., 1955; Wightman & Kistler, 1992; Zurek, 1993).

**Significance:** This information is helpful in designing future digital hearing aids and for counselling clients for whom a hearing aid is part of the rehabilitation program.

**References:**


Gain mismatch and localisation performance

Investigators: Gitte Keidser, Elizabeth Convery, Harvey Dillon, and Volkmar Hamacher

1 Siemens Hearing Instruments, Germany

This study is supported by CRC Hear.

Brief report: There is a potential risk that bilaterally fitted hearing aid users with access to a volume control may adjust the gain on each device in a way that results in an imbalance in loudness across ears. A loudness mismatch across devices can interfere with the interaural intensity difference (IID) cue that contributes to discrimination between sound sources located to the left or right of the median plane and hence localisation performance in the horizontal plane. Nine subjects with symmetrical hearing loss were binaurally fitted with a pair of Triano S behind-the-ear (BTE) devices according to the NAL-NL1 prescription. After establishment of the overall gain in each device that provided a perfect loudness balance across ears, the subject was fitted with 3, 6, and 9 dB gain mismatch between ears relative to the 0 dB gain mismatch condition. Each subject then completed horizontal localisation testing (see description of localisation array and test procedure elsewhere in this report) using five stimuli with different spectral shape: wideband speech, wideband low-frequency weighted noise (traffic), wideband high-frequency weighted noise (cockatoos), narrowband low-frequency weighted noise (one-octave 400 Hz pink noise), and narrowband high-frequency weighted noise (one-octave 3150 Hz pink noise).

From the subjects’ responses to the localisation test, a total RMS localisation error in degrees and a modified RMS localisation error in degrees that ignores front/rear confusions was calculated. The RMS errors revealed no significant effect of gain mismatch (p > 0.3). In addition, for each subject, the mean actual error score in degrees when ignoring front/rear confusions was calculated. This error score showed that the subjects, when listening to the one-octave 3150 Hz pink noise, increasingly directed their responses towards the side where the intensity increased as the gain mismatch increased, see Figure 1.

This finding was statistically significant, p < 0.03. The one-octave 3150 Hz pink noise was the only stimulus that did not contain low-frequency information for which the interaural time difference (ITD) cue is dominant (Wightman and Kistler, 1992).

Figure 1. The average actual modified errors (front/rear confusions ignored) measured for each stimulus and gain mismatch.

Reference:

The advantages of wide-dynamic-range compression over linear amplification for children

Investigators: Teresa YC Ching, Mandy Hill, Emma van Wanrooy, Katrina Agung

Background: People with sensorineural hearing loss usually have a reduced dynamic range from just audible level to uncomfortable listening level, but sound levels in the environment vary over a very wide range. With linear amplification, it is not always possible to restore audibility to weak sounds without intense sounds being over-amplified. This problem can be alleviated by the use of automatic gain control, which can be implemented in various ways for different purposes (see Dillon, 1996 for a tutorial). Compression limiting systems can be used to limit the maximum output of hearing aids. The output increases linearly with increase in input level up to a certain point. After that, the output is held constant by circuitry that automatically turns down the gain of the hearing aid by one decibel for every decibel that the input level increases. This provides listening comfort while reducing distortion associated with peak-clipping methods for output limiting. Wide-dynamic-range compression (WDRC) systems that allow automatic gain control to begin at low sound input levels enable signals that vary in level over a wide range to be amplified as a narrow range of output signals. This method of amplification increases audibility of weak sounds while keeping intense sounds at a comfortable listening level, and is expected to lead to improved speech perception for low-level input sounds and improved comfort for high-level input sounds. Non-linear amplification, particularly when it incorporates compression with a low compression ratio, has advantages for listener comfort and speech understanding over linear amplification for most people with hearing loss.

These advantages are likely to be even greater for children (Dillon, 2000), because they are unable to turn up the hearing aid gain in order to hear soft sounds well for speech intelligibility, and to turn down the gain when loud sounds become too loud for listening comfort. A hearing aid with WDRC does this automatically. However, non-linear amplification has not been commonly used for children in clinical practice (Tharpe et al, 2001).

The empirical evidence on whether WDRC provides more benefits over linear amplification for children is not well established. In general, all studies agreed that the use of WDRC did not lead to any decrement in speech scores compared to linear amplification. All agreed that the WDRC did not provide speech benefits over linear amplification for speech presented at an average input level. However, conflicting results are reported for speech presented at low and high input levels, and for speech presented in noise. Two studies found a significant improvement with WDRC over linear amplification at low and high speech levels (Jenstad et al, 1999; Gabbard et al, 2003) but one did not (Marriage & Moore, 2002). Benefits were reported for speech perception in noise in some studies (Bamford et al, 1999; Gabbard et al, 2003) but not others (Marriage & Moore, 2002). The divergent findings may be related to differences in methodology and effects of uncontrolled variables that might have confounded the outcomes. For instance, when different hearing aid circuitries and fitting algorithms were used for linear and WDRC amplification (Bamford et al, 1999; Gabbard et al, 2003), it is not clear to what extent a difference in speech scores was due to the difference between WDRC and linear amplification. Only one study allowed for the effects of acclimatization. None of the evaluations were carried out using a double-blind design.

This project aimed to determine the relative effectiveness of linear and WDRC amplification for children with mild to severe hearing loss using a double-blind, cross-over design that allowed...
for acclimatization. Effectiveness was assessed by using outcomes measures that included speech perception, loudness rating, horizontal localization, and functional performance in real life.

**Research question and hypotheses:** Is wide-dynamic-range compression more beneficial than linear amplification with compression limiting for children with mild to severe degrees of hearing loss? It was hypothesized that:

1. Children could hear soft speech better with WDRC than with linear amplification, regardless of the degree of hearing loss.
2. Children rated loud speech to be more comfortable when listening with WDRC than with linear amplification, regardless of the degree of hearing loss.
3. Children functioned more effectively in everyday life when using WDRC than linear amplification, regardless of the degree of hearing loss.
4. Children preferred WDRC to linear amplification in everyday life situations, regardless of the degree of hearing loss.

**Procedure:** Twenty children with bilateral mild to severe hearing loss participated in this study. Their age ranged from 6-18 years. The duration of hearing aid use ranged between 3 to 16 years.

Each child was fitted bilaterally with new behind-the-ear non-linear hearing aids according to the NAL-NL1 prescription (Dillon, 1999). The real-ear-to-coupler difference for each individual was used to derive coupler gain targets that matched real-ear aided gain targets of the prescription (Ching et al, 2002). Hearing aids were adjusted in an HA2-2 cc coupler to verify that prescribed targets were achieved.

The two options compared were linear (LIN) and wide-dynamic-range-compression (WDRC) amplification. In the linear mode, the compression parameters were deactivated, but compression limiting was used to limit the maximum output of the hearing aid. The gain-frequency response for an average input level would be the same for the LIN and WDRC programs, but the gain-frequency response for low and high input levels would differ. Half of the children wore hearing aids set to WDRC first, and the other half to LIN first.

There were five home-trial periods. Neither the audiologist who conducted the tests nor the child/parent knew which circuit was being used during each period. During the first two periods, the children wore hearing aids set to each mode for 4 – 8 weeks. During the third home trial period of 2 – 4 weeks, the children could switch between linear and non-linear programmes. The programmes were swapped for the fourth home trial period of 2 – 4 weeks. Finally, the volume control of the hearing aids was activated at the fifth home trial period, and the children could switch between programmes as well as adjust the volume control for each programme if desired. At the end of each home trial period, speech perception, localization, and loudness rating were evaluated. Functional performance in everyday life was assessed by using a parent’s questionnaire. For the third and fourth home trial periods, the children were required to complete a diary to document the performance of different options in a range of everyday life situations. For the fifth home trial period, the children were asked to record any volume control adjustments required for either or both programmes.

**Speech perception:** The BKB sentence test material was used to measure speech reception thresholds of sentences presented at 55 and 70 dB SPL in speech-shaped noise. The speech and noise signals were presented from a loudspeaker located 1 metre from the subject position at 0 degrees. Performance was measured in terms of signal-to-noise ratio for 50% correct identification for each level. A nonsense syllable test (NAL CD) was used to measure identification of consonants presented at 55, 70, and 80 dB SPL in quiet. Performance was scored as percent correct identification.
Localization: Horizontal localization was evaluated by using an array of 11 loudspeakers spanning 180 degrees located in an anechoic chamber. Pink noise pulses were presented in a random order from one of the loudspeakers, and the children were asked to identify which loudspeaker the noise came from. Performance was measured in terms of error (in degrees) between the source loudspeaker and the response loudspeaker.

Loudness rating: Sentences were presented at levels ranging from 55 to 80 dB SPL in 5 dB steps. The children were asked to rate the loudness on a 7-point scale that ranged from “Can’t hear” to “Much too loud”.

Functional performance: For the first two home trials, the parents were asked to observe their child’s functioning in everyday life. Their observations were systematically elicited and recorded during an interview based on a questionnaire. The questionnaire comprised 3 questions on usage of device, 5 questions on aural/oral communicative function in quiet situations, 4 questions on communicative function in noisy situations, and 2 questions on environmental awareness. Subscale scores on use, quiet, noise, and environmental awareness were obtained by averaging scores across questions for each category, and scores for all questions were summed to yield an overall score. For the third and fourth home trials, the children were given a diary in which they recorded their overall preference and rated their preference for a range of everyday life situations. Thirteen children completed the diaries. An overall preference rating and an averaged rating across all situations were scored.

Analysis of variance with repeated measures was used to determine whether or not significant differences existed between the linear and the non-linear conditions for each outcomes measure. When the overall p-value was significant (p < 0.05), multiple post-hoc testing with the Tukey’s Honest Significance Difference test and Bonferroni adjustment of p = 0.02 was used to define significant comparisons for the two treatment conditions.

Preliminary analysis: This project is currently in progress. The following reports a preliminary analysis of results from 14 children.

The left panel in Figure 1 shows the averaged low-frequency (250 – 1000 Hz) gain for the linear programme compared to the non-linear programme (WDRC). Data points above the diagonal line show that more gain was provided by the non-linear programme compared to the linear programme. The right panel shows that the same is observed for the averaged high-frequency (2000 – 4000 Hz) gain.

![Figure 1](image.png)

Figure 1. This figure shows the averaged gain provided by WDRC in relation to that provided by linear amplification for an input level at 55 dB.

Loudness rating: Both LIN and WDRC mapped 55 to 80 dB speech to a range between “too soft” and “too loud”. Loudness growth was not significantly different between linear and WDRC, and both were not significantly different from the normal loudness growth function. The following figure shows the average loudness rating of hearing-impaired children when they used the linear program (circles) compared to the WDRC program (squares). Data from a group of normally hearing young adults are shown by the filled triangles.
Figure 2. Loudness rating averaged across 14 children for the linear programme (circles) and the non-linear programme (squares). Vertical bars denote 0.95 confidence intervals.

Speech perception: Sentence perception in noise was equally good between the two amplification options, both for speech presented at 55 and 70 dB SPL. Consonant perception in quiet was similarly good at medium and high input levels. However, the consonant scores for 55 dB input were significantly higher for WDRC than for linear amplification (p < 0.05).

Figure 3. Mean consonant scores for linear (square) compared to non-linear (circles) amplification for speech presented at 55, 70, and 80 dB SPL. Vertical bars denote 0.95 confidence intervals.

Horizontal localization: Figure 4 shows the median rms error (degrees) for localization of 9 children at presentation levels of 55, 70, and 80 dB SPL. ‘L’ denotes linear, and ‘NL’ denotes non-linear. In the linear condition, errors reduced with increase in presentation level. This suggests that increased audibility assisted in improving the children’s ability to localize the source of sounds. In the WDRC condition, errors were similarly low across all three input levels, and the order of magnitude is equivalent to that of the linear condition for an 80 dB input.

Analysis of variance using error scores for all conditions as dependent variables, program (linear vs non-linear) and level (55, 70, 80 dB) as repeated measures factors showed that the interaction between program and level was significant (p < 0.05). Post-hoc analyses revealed that localisation at low input level was significantly better with WDRC than with linear amplification (p < 0.04).

Figure 4. Horizontal localization errors.

Functional evaluation: The parents rated the children’s performance with WDRC to be superior to the linear program (p < 0.05).

Figure 5 shows the averaged scores for the four subscales (Use, communicative functioning in quiet situations, in noisy situations, and environmental awareness), based on the parents’ ratings.
Figure 5. Parents’ evaluation of children’s functional performance in everyday life when the children wore hearing aids set to linear (circles) compared to WDRC (squares).

Analysis of variance using program (linear vs non-linear) and scale (use, quiet, noise, environmental awareness) as repeated measures factor indicated that the main effects of program and scale were significant (p < 0.003 and p < 0.00001 respectively). The interaction was significant (p < 0.001). Post-hoc analysis revealed that perception of environmental sounds was significantly better with WDRC than with linear amplification (p < 0.0003). Ten of the thirteen children who completed the diaries consistently preferred the WDRC program in a range of communicative situations in real life, and the other three did not have a consistent preference.

In summary, the preliminary results demonstrate that children with moderate hearing loss benefited from the use of WDRC due to increased audibility for soft input levels. They recognized soft speech better and localized sounds better with WDRC than with linear amplification. It remains to be seen whether the same conclusions apply to children with more severe hearing loss.

Significance: This double-blind study clearly demonstrates the benefit of WDRC over linear amplification for children with moderate degrees of hearing loss. Children should be fitted with non-linear amplification wherever possible.

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Reliability of client-based adjustments of amplification

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Conducted as part of CRC HEAR.

Background: The fitting of modern hearing aids is becoming increasingly complex. An important reason for this trend is the fact that modern hearing aids aim to compensate for more than the audibility loss. Techniques have been introduced that try to compensate for supra-threshold deficits, using non-linear approaches in multiple bands, and that either enhance the speech signal or reduce the amount of background noise. Fitting rules for the techniques in this area are not available yet and fitting is guided by trial and error or by proprietary fitting rules for specific devices that are usually not evidence-based and mostly not properly documented.

Future developments may facilitate an alternative approach to fitting where the subject will be allowed to control the most important fitting parameters in his/her own acoustical environment. The concept of a trainable hearing aid is promising in this respect. If technology allows, hearing aid users may be equipped with a training unit for the first weeks of use and/or for a change of the hearing aid setting when the hearing loss has changed and/or the most common acoustical environments have changed (e.g. after job rotation). Such a concept may circumvent a number of the disadvantages in the current fitting approaches as it includes the user’s personal preferences and specific listening situations into the fine-tuning process of the hearing aid. Important prerequisites of such approach are that the hearing aid user is able to make consistent adjustments, and that the individually fitted solutions are significantly better accepted than a set of hearing aid settings according to some fixed fitting rule.

This study was carried out at the Bionic Ear Institute and at NAL as part of a research line within the Cooperative Research Centre for Hearing Aid and Cochlear Implant Innovation to develop a trainable hearing aid and to test a number of aspects related to the applicability of the concept. In this study the listener is given access to a set of gain controls and while listening to selective sound samples that represent real-life listening situations the client is able to adjust the amplification characteristics to the preferred settings. The experimental questions of this research are:

1. Is the hearing aid user able to provide systematic and reproducible control adjustment with respect to his preferences in different acoustical conditions, in this study simulated in a laboratory set-up?
2. Given a limited number of control functions, which of a set of four different controllers is most effective, based on subject preference and test-retest reliability?
3. How does test-retest reproducibility compare to the differences in amplification needs within a limited set of acoustical environments, used in this experiment?

Procedures: In a laboratory setting, hearing-impaired subjects were given access to different ways to fine tune the frequency-gain curves according to their individual preferences, using four different controller configurations labelled A, B, C, and D. Controller A enabled the subject to adjust the volume and the slope of the response. Controller B enabled the subject to adjust the volume, the slope, and the mid-frequency contrast. With controller C, the subjects could adjust gain independently in the low, mid, and high frequencies, whereas the volume and gain in the low and high frequencies were adjustable with controller D. Gain changes in response to a key press were 2 dB. We used a fixed set of acoustical environments, presented to the subjects by video segments. The video segments were presented in a free-field condition using a TV-screen and two...
high-quality loudspeakers, that were situated 2.5 m in front of the listener (see Figure 1). The listeners were seated in a large sound-insulated room and did not use their hearing aids to avoid uncertainty from specific effects of the signal processing in the hearing aids or acoustical effects in the individual ear moulds. Four different control configurations were tested in four different test sessions.

After training, the control configuration under test was used to fine tune the amplification settings for the six videos, starting from 4, 8, 12, or 16 dB overall shallower or steeper responses than the NAL-RP prescription both in test and retest (6 x 2 x 2 = 24 settings). We applied a roving procedure to the gain and frequency response slope of the baseline characteristic (used as starting response) and the order of test conditions was balanced according to a Latin square. Each test session ended with a questionnaire on the subjective judgements of the subjects relating to a number of aspects of the controller under test. The order of the control configurations across sessions was also balanced according to another Latin square.

At the end of the final test session, some additional questions were answered relating to the subjects' preferences across controllers and his/her interests in applying such controllers in his/her own hearing aid.

Results:

1. The hearing aid user is able to provide reproducible results. With respect to reproducibility, the test-retest standard deviations of gain at each frequency are 2.9 dB, averaged across subjects, frequencies, videos, and baseline configurations. One of the controllers (controller B, using a volume key and two tone balance keys) was clearly worse (>3.6 dB). For the other controllers the test-retest standard deviation was clearly better (about 2.5 dB).

2. The hearing aid user is satisfied about the use of one or more of the controllers. The subjective results derived from the questionnaires are very favourable with respect to the ease of use and the degree to which the subjects could improve sound quality, speech clarity, and sound comfort. One aspect that probably can be improved further is the size of the perceptual change corresponding to each key press. Figure 2 shows that a number of step sizes was judged to be “somewhat subtle”, especially for controller C (using 3 keys for bass, mid, and treble). This is probably because controller C did not have a volume control and hence a change in overall gain required three key presses compared to one key press on the other controllers.

3. Controllers A and D were judged as the most appropriate. The final questionnaire showed that 12 subjects (50%) preferred controller A (using 2
keys for volume and slope) and 9 subjects (37%) preferred controller D (using 3 keys for volume, bass, and treble). These subjective results correspond well with more objective parameters like the poor reproducibility for controller B (see above) and the speed of adjustment. Controllers A and D needed, on average, about 13 key actions, while controllers B and C needed more than 17 key actions per fine tuning.

4. The test-retest reproducibility is high enough to assess systematic differences in amplification for the limited set of acoustical environments, used in this experiment.

For each of the controllers, the reproducibility was high enough to produce significant differences for the main effects of subjects and videos. Unexpectedly, also the effect of the baseline configuration was significant.

5. The starting configuration determines the final gain preference to a large degree. Figure 3 shows the final settings for the overall gain (y-axis, relative to the individual NAL-RP prescription) for each of the six videos (along the x-axis). The pattern across videos illustrates the consistent differences in terms of overall gain preference between different videos. The four curves correspond to different (roved) levels of the overall gain of the baseline configuration used as starting configurations. The horizontal shifts of these curves indicate a certain degree of conservatism in fine tuning: starting confirmations with less overall gain (marked as -6 and -4) also result in final settings with less gain.

Likewise, Figure 4 shows the final settings for the gain slope (y-axis, relative to the slope of the individual NAL-RP prescription) for each of the six videos (along the x-axis). The seven curves correspond to different (roved) levels of the gain slope of the baseline configuration used as starting configurations. This figure shows that a start at “steeper than NAL” (marked as 8, 6, 4, and 2) gives the steeper amplification curves and a start at “shallower than NAL” (marked as -8, -6, and -4) gives the flatter amplification curves. The degree of roving matches almost perfectly with the order of the gain slopes finally preferred.

Significance: This study is just one step in testing the applicability of a trainable hearing aid. The study is a laboratory study with a limited number of environments and a limited number of fitting parameters. However, the results are encouraging. The subjects produce reproducible results with clear differences between specific acoustical environments. This implies that consistent differences across environments are wanted and the trainable hearing...
aid allows the subjects to arrive at their preferred hearing aid settings for each individual listening environment. Most importantly, the subjects are able to handle different control configurations. This suggests that the control parameters may be extended towards more complex hearing aid functions without problems.

Even within the limitations of a laboratory study, the subjects received a reasonable impression about the possibilities of controlling loudness and the response shape in a real-life application. In the final questionnaire we asked how likely it would be that the subject would use such a controller in real-life listening situations. Figure 5 shows the results and indicates that the majority of subjects would like to use a controller like the one tested in real life.

One unexpected outcome was the large impact of the starting condition (baseline) on the preferred frequency response. This finding implies that within one session only a biased correction of optimal amplification can be reached. We have to assume that this probably complicates each hearing aid fitting, also in current clinical practice. The “real optimum” may be expected to be reached after successive steps in more than one session and the most practical way to achieve that is in a daily “training program”. So, also from this perspective, the concept of a trainable hearing aid is worth to be investigated in further detail.

Factors affecting speech intelligibility of hearing-impaired listeners

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Conducted as part of CRC HEAR.

Background: Many hearing-impaired people have difficulty understanding speech even though they can hear the signal with amplification. As hearing loss increases, detection of voices and other sounds would still be possible when the sounds are intense enough, but extraction of speech information from the audible signal would be much reduced. This is referred to as ‘hearing loss desensitization’ (Pavlovic et al, 1986; Studebaker & Sherbecoe, 1992; Studebaker et al, 1997). Desensitization is a key issue in prescribing hearing aid amplification that aims to maximise speech intelligibility. This is because it implies that more gain should be provided in regions where hearing is less impaired than where it is more than would otherwise be the case were all regions to be equally effective.

The empirical specification of desensitization as a function of hearing loss at different frequencies makes it possible to derive the best combination of gains across frequencies that optimises speech intelligibility (Ching et al, 2001), as implemented in the NAL-NL1 prescription (Dillon, 1999). Given that the approach was based on data averaged across listeners, gains prescribed might be inappropriate for individuals who had better- or poorer-than-average abilities to extract speech information. For this reason, attempts were made to explain individual desensitization in terms of frequency and temporal resolution, in addition to hearing sensitivity. Although poorer resolution must be associated with lesser proficiency at extracting speech information from an audible signal, both are also directly linked with greater hearing loss. After
controlling for the average desensitization based on hearing loss, Ching et al (1997; 2002) found that the better- or poorer-than-average resolution for a certain degree of hearing loss accounted for a very small proportion of variance in individual desensitization. The issue remains as to whether additional factors that significantly account for why two persons presenting with similar audiograms extract different amounts of speech information from an audible signal can be identified, so that hearing aids can be prescribed appropriately for the individual listener.

Hearing loss desensitization appears to be greater at high than at lower frequencies (Ching et al, 1998; Hogan & Turner, 1998; Amos et al, 2000). Recent research suggests that some people do not benefit from amplification in the high frequencies because of the presence of dead regions, or regions with no functioning inner hair cells and/or auditory neurons, in the cochlea (Moore, 2001, 2004; Vickers et al, 2001; Baer et al, 2002). When a listener has a cochlear dead region, the audiogram does not reveal the amount of hearing loss in that region (Moore, 2001; Halpin et al, 1994; Halpin, 2002). The listener will respond to a test tone whose frequency falls in the region if the tone is intense enough to stimulate the nearest region that contains functioning hair cells. This is referred to as ‘off-frequency listening’ (Gravendeel & Plomp, 1960; Halpin et al, 1994). The frequencies in a speech signal that fall into cochlear dead regions will still be audible, but the information contained therein will not be processed adequately because it falls in a region tuned to other frequencies (Shannon et al, 1998). It may even interfere with extraction of information contained in frequencies corresponding to the functional region. This may explain why some people with severe sloping hearing loss performed more poorly in speech tests when more high-frequency amplification was provided (Turner & Cummings, 1999; Ching et al, 2002). However, the presence of cochlear dead regions was not directly measured in these studies.

The evidence from Vickers et al (2001) and Baer et al (2002) shed new light on the plausible connection between cochlear dead regions and desensitization. They tested subjects for the presence of cochlear dead regions by measuring masked thresholds in a threshold equalizing noise (TEN, Moore, 2001) and related the results to the consonant test scores obtained with low-pass filtered speech presented in quiet and in noise. They showed that on average, speech scores of subjects with no dead regions increased with widening of bandwidth up to 7500 Hz. For the 6 ears (5 subjects) with cochlear dead regions, test scores tended to increase with cut-off frequencies above the edge frequency of the dead regions, then remained nearly constant or decreased slightly when the cut-off frequency exceeded one octave of the estimated edge frequency of the dead region. In these studies, subjects with cochlear dead regions also had greater high-frequency loss than subjects without dead regions.

Would audibility have explained speech intelligibility of subjects with cochlear dead regions? This question can be evaluated by adopting the framework of the Articulation Index model (Fletcher, 1953; ANSI Standard, 1969). The model makes it possible to estimate speech scores from the amount of speech signal that is above the listener's hearing threshold or noise, whichever is higher. Baer et al (2002) used a modified Articulation Index that takes into account the broadened filter bandwidth of hearing-impaired listeners (Moore & Glasberg, 1998), and showed that the consonant test scores obtained by the 5 subjects with dead regions were below the scores predicted on the basis of audibility. They concluded that the presence of dead regions explained the overestimation. In contrast, reanalysis of the data from Vickers et al (2001) by Rankovic (2002) revealed that the Articulation Index model of Fletcher (Fletcher, 1953) adequately predicted speech scores for subjects with or without dead regions. In a separate study, Vestergaard (2003) used the TEN test to identify cochlear dead regions and...
word test material low-pass filtered at different cut-off frequencies to measure speech recognition ability of 10 adults listening with their hearing aids. It was found that the Standard SII method (ANSI, 1997) adequately predicted the performance of subjects with or without cochlear dead regions. For the same restricted amount of audibility, it was also found that the subjects with cochlear dead regions (who also had more steeply sloping hearing loss) performed better than those without cochlear dead regions.

Whether hearing loss desensitization has a greater effect on older than on younger people is also controversial. Whereas some previous research suggested that audibility accounted for speech intelligibility of older people (Van Rooij & Plomp, 1992; Humes et al, 1994), recent research identified age (greater than 70 years) as a significant factor after controlling for audibility (Studebaker et al, 1997). Humes (2002) investigated this issue by assessing the speech recognition performance as well as the auditory processing and cognitive ability of a group of 171 adults who were aged between 60 and 89 years. It was found that speech performance was over-estimated by the SII model even after the hearing loss desensitization proposed by Studebaker et al (1997) was included. Stepwise linear regression analyses using audiological measures, auditory processing measures, and cognitive measures as predictor variables and speech recognition as dependent variable showed that although audibility was the single best predictor of the speech performance, age-related changes in cognitive function also accounted for some of the variations in performance.

Adopting this method in reanalysing data from three previous studies, Humes (2003) concluded that speech recognition performance of older hearing-impaired people would be best predicted by hearing loss, cognitive performance, and age. These findings suggest that cognitive ability and/or age may contribute to explaining why some hearing-impaired listeners cannot extract as much speech information from an audible signal as other listeners with the same degree of hearing loss.

To summarise, whereas there is a general acceptance that the ability to extract speech information from an audible signal decreases with increasing hearing loss and age, the underlying factors affecting this ability are not well understood. The broad objective of this project was to investigate factors affecting the effectiveness of audibility for speech understanding by people with different degrees of hearing loss.

Research question and hypotheses: How much of the variability in performance on speech tests can be predicted from outer hair cell function, the presence/absence of cochlear dead regions, psychophysical tuning curves, cognitive ability, and age, after controlling for the effect of hearing threshold levels?

It was hypothesized that for the same degree of hearing loss,
1. people without cochlear dead regions would be more proficient in extracting speech information from a certain amount of audibility than people with cochlear dead regions;
2. people with sharper tuning in a frequency region would extract more speech information from the same amount of audibility than people with poor tuning;
3. people with greater cognitive ability would extract more speech information from the same amount of audibility than people with lesser ability;
4. people who are younger would extract more speech information from the same amount of audibility than people who are older.

Procedure: Twenty-two normally hearing adults and fifty-three hearing-impaired adults with hearing loss ranging from mild to profound degrees participated in this study. Their age ranged from 20 to 86 years (median = 70 years). Outer hair cell function was assessed using transient-evoked oto-acoustic emissions. Cochlear dead regions were assessed at 250, 500, 1000, 1500, 2000, 3000, 4000, 6000,
and 8000 Hz by using the TEN test (Moore, 2001). Psychophysical tuning curves were measured at 500, 1000, 2000, and 4000 Hz by using a pulsed tone as signal and narrow-band noise at different frequencies as maskers. Cognitive ability was measured using a visual digit-monitoring task (Knutson et al, 1991) and a visual letter-monitoring task (MRC Institute of Hearing Research, UK).

The ability to extract speech information from an audible signal at different frequencies was assessed by using filtered speech materials presented in quiet and in babble noise. The speech materials included sentences (City University of New York sentence lists) and consonants (VCV test) high-pass (HP) and low-pass (LP) filtered at 700, 1400, and 2800 Hz; and low-pass filtered at 5600 Hz (the conditions are abbreviated as: LP700, LP1400, LP2800, LP5600, HP700, HP1400, HP2800). The speech stimuli were individually shaped and amplified according to the POGO prescription (Schwartz, Lyregaard, & Lundh, 1988), and presented monaurally via Sennheiser HD-25 SP earphones.

**Preliminary analysis:** The Speech Intelligibility Index model (ANSI S3.7, 1997) was used to relate speech scores to audibility. The results of normal hearing listeners were used to derive transfer functions by using a least squares procedure. Because there was considerable spread of data across filter conditions, with scores for the narrower bands consistently lower than those for the broader bands, transfer functions were separately derived for speech scores of each filter condition, both in quiet and in noise. The following figure (Figure 1) gives the set of transfer functions for the VCV test.

**Figure 1.** Different transfer functions relating VCV scores to calculated SII.

The speech scores of hearing-impaired listeners were related to the transfer function for each filter band. Individual proficiency for each band was

**Figure 2** gives an example of the calculation of desensitization.

**Figure 2**

**Figure 3.** PTC Q10 value as a function of hearing threshold level at 2000 Hz.
expressed as a ratio of the apparent SII (SA) based on the speech score relative to the SII calculated from the hearing thresholds, speech and noise levels measured at each one-third octave band (S). **Figure 2** illustrates how proficiency was calculated. The curve is the transfer function relating speech scores to the Speech Intelligibility Index. The score of an individual participant when calculated audibility (S) is 0.6 is represented by a circle, the apparent SII (SA) is 0.3. This gives a proficiency of 0.5, implying that the participant extracted about half of the speech information that is available to the normal-hearing person. Proficiency values that are less than one denote desensitization.

The NAL-OAE software was used to analyse the otoacoustic emissions data to yield an index of coherent emissions strength (CES) for each test ear. From the TEN test, the elevation of the masked thresholds at each measured frequency was quantified by taking the difference between the masked threshold and the TEN level or the threshold in quiet, whichever was higher. An elevation of greater than 10 dB denotes the presence of a cochlear dead region at the frequency of measurement. The PTCs were analysed in terms of sharpness of tuning, represented by the bandwidth of the tuning curve at 10 dB above the level of the masker at the signal frequency ($f_s$) or the $Q_{10}$ value, specified as $Q_{10} = f_{ip} / \text{bandwidth}$, where $f_{ip}$ is the signal frequency. A high $Q_{10}$ value denotes sharp tuning, and a value of zero indicates that the bandwidth is unquantifiable due to the flatness of the tuning curve.

The location of the PTC tip was specified as $\text{Shift} = \log_2 (f_{ip}/f_{ip})$, where $f_{ip}$ is the frequency at the tip. Any value of Shift that deviates from zero represents a shift in tip frequency measured in octaves. Data on cognitive ability are currently being collected.

This report presents preliminary analyses in relating the derived proficiency of each filtered speech band presented in quiet and in noise to averaged hearing thresholds and PTC $Q_{10}$ values for the band.

**Figure 3** shows the PTC $Q_{10}$ values as a function of hearing level at 2000 Hz. Open circles represent results from subjects with PTC tips centred on the signal frequency, and filled triangles represent those with PTC tips shifted to a neighbouring masking frequency.

The band proficiency at LP2800 was negatively correlated with averaged HTL ($r = -0.35$ and $-0.61$ for proficiency in quiet and in noise respectively, and $-0.45$ overall), but was positively correlated with PTC $Q_{10}$ ($r = 0.36$ and 0.64 for proficiency in quiet and in noise respectively, and 0.45 overall). **Figure 4** shows the deterioration of proficiency for the low-pass filtered 2800 Hz band (LP2800) with increasing hearing loss in the left panel, and with lower $Q_{10}$ or poorer tuning, in the right panel. The regression of proficiency in quiet (solid line) and proficiency in noise (broken line) with hearing threshold (left panel) and $Q_{10}$ (right panel) are shown.
Forward stepwise linear regression analysis was carried out to identify which of the factors, hearing loss, PTCQ_{10}, presence/absence of cochlear dead regions, age, or combination of factors, contributed significantly to the proficiency of each filter band. Band proficiency was the dependent variable, and OAE strength, PTC Shift, and TEN elevation at the highest measured frequency in each filter band, hearing threshold levels averaged across the band, and age were independent variables. Significant results are summarised in Table 1 below. The results suggest that age and hearing threshold levels are the primary predictors for proficiency, and that frequency resolution (PTC or TEN) or OAE strength do not account significantly for further variance.

![Table 1](image)

**Significance:** Preliminary analyses confirm that for the same amount of audibility, the proficiency to extract speech information decreases with increasing hearing loss, both for listening to speech in quiet and in noise. Proficiency also decreases with decreasing frequency selectivity as measured by PTC. Frequency selectivity decreases with increasing hearing loss and with age. Forward stepwise linear regression analyses show that once audibility has been allowed for, age is the best predictor for speech intelligibility, followed by hearing threshold levels. The extent to which cognitive ability affects speech intelligibility after accounting for hearing loss remains to be investigated. Further analyses will be carried out to examine 1) the potential usefulness of measures in addition to hearing thresholds for predicting how much speech information a hearing-impaired person can extract from an audible signal, 2) the likelihood that incorporation of the measure(s) would lead to a different prescription, and 3) the possibility that such a prescription would lead to better speech intelligibility than one that is based exclusively on hearing thresholds. Resolving these issues is crucial to prescribing amplification that maximizes speech intelligibility for hearing-impaired listeners.

**References**


Diagnosing dead regions in the cochlea: PTC or TEN?

Investigators: Teresa YC Ching, Harvey Dillon, Frances Lockhart, Emma van Wanrooy, Lydia Lai

Many studies have shown that when hearing loss is greater than 60 dB at high frequencies, amplification of these frequencies may increase audibility but not speech intelligibility (Amos & Humes, 2001; Ching et al, 1998; Hogan & Turner, 1998; Rankovic, 1991; Turner & Cummings, 1999). Moore (2001) posited that the key to the degree of speech benefit due to high-frequency amplification may be the presence or absence of ‘dead regions’ in the cochlea. A cochlear dead region is a region in the cochlea where inner hair cells and/or neurons are functioning so poorly that a tone producing peak basilar-membrane vibration in that region is detected by off-place listening (Moore 2004). Based on the data from Vickers et al (2001) and Baer et al (2002) showing an increase of speech scores with greater audibility for subjects with no cochlear dead regions but no increase for those with dead regions, Moore (2001; 2004) proposed that there is little or no benefit from amplification beyond 1.7 – 2 times the edge frequency of a cochlear dead region. However, researchers do not agree on whether dead region considerations are useful for determining amplification needs (Rankovic, 2002; Moore, 2002).

Identification of cochlear dead regions may be important for prescribing amplification only if considering the presence of dead regions in addition to an audiogram leads to an alternative prescription, and that the prescription results in better speech intelligibility than a prescription that is based on the audiogram. For practical purposes, this can be broken into a) whether the presence of dead regions can be reliably determined using current methods; and b) whether the presence of dead regions explains speech recognition deficits better than the audiogram alone. This report focuses on the first of these questions by reviewing current literature and by evaluating the...
relative efficacy of methods that have been widely used for determining cochlear dead regions.

Two methods used for diagnosing cochlear dead regions are the measurement of psychophysical tuning curves (PTC) (Florentine & Houtsma, 1983; Turner et al, 1983; Moore & Alcantara, 2001) and the tone detection in threshold equalising noise (TEN) tests (Moore, 2001; 2004). To measure a PTC, the signal (usually a pure tone) is set to a level just above the listener’s threshold, say 10 dB SL. The masker can be either a pure tone or a narrowband noise. For each of several masker centre frequencies, the level of the masker needed to just mask the signal is determined. When there is no dead region, the tip of the PTC, or the frequency at which the masker level is lowest, should lie close to the signal frequency. When there is a dead region at the frequency of the signal, the tip of the PTC should shift to the nearest region with surviving hair cells, because the masker centred at that region should produce the most effective masking. In other words, a shifted tip is indicative of a dead region, and the frequency at the tip of the PTC can be used to define approximately one boundary of the dead region (Moore & Alcantara, 2001).

The interpretation of the PTC is not without complications (Moore, 2004; Kluk & Moore, submitted). Notably, the shapes of PTCs for hearing-impaired listeners can be influenced by the detection of beats and simple difference tones thereby giving a misleading impression that the PTCs have tips at the signal frequency, even when the signal frequency falls in a dead region.

An alternative method proposed by Moore (2001) for diagnosing dead regions is based on measuring the level for detecting pure tones in a ‘threshold equalising noise’ (TEN). This noise is spectrally shaped to produce equal masked thresholds at all frequencies (250 – 10,000 Hz) in dB SPL, and the test is referred to as the TEN test. When the TEN test is administered to a listener with no cochlear dead regions, the masked thresholds are within 10 dB of the TEN level/ERB (equivalent rectangular bandwidth). Masked thresholds exceeding 10 dB of the TEN level, with the condition that the TEN produces at least 10 dB masking, are indicative of a dead region. These criteria were developed using a small group of older adults with moderate to severe cochlear hearing loss (Moore et al, 2000), and may need to be revised for young people or people with more severe hearing loss (Moore, 2004).

Despite the relative ease of administration, inconclusive, rather than positive or negative results, are frequent when the test is applied to people with severe or profound hearing loss (Moore et al, 2003). Either the TEN level is not intense enough to provide at least 10 dB of masking, or the noise level is not intense enough to permit the measurement of a masked threshold without causing loudness discomfort. Although it is likely that a cochlear dead region is present when the hearing loss is more than 90 dB HL in the high frequencies (Moore, 2001), the TEN test cannot be applied adequately in these cases due to loudness limitations. A more recent version of the test, the TEN (HL) test, uses a bandlimited TEN designed to produce equal masked thresholds at frequencies between 500 and 4000 Hz in dB HL, thereby enabling the test to be used with hearing loss up to 100 dB HL (Moore et al, 2004). A fast PTC measurement technique has been proposed for defining the edge frequency of dead regions after a diagnosis of dead regions was made using the TEN test (Sek et al, submitted).

Although both the PTC and TEN tests have been applied to identifying cochlear dead regions in recent research, the results are controversial with one study lending support to the TEN test (Moore et al, 2000) whereas a second study queries the reliability of the TEN test (Summers et al, 2004).

This report addresses the question of whether cochlear dead regions can be reliably determined by examining
the consistency between the PTC and TEN tests in 53 hearing impaired listeners. The related issues of whether the presence of cochlear dead regions determines the contribution of audibility for speech intelligibility; and the usefulness of knowledge about dead regions for prescribing amplification will be examined in a separate report.

**Research question and hypotheses:**

Do PTC and TEN tests give consistent results in identifying cochlear dead regions in hearing-impaired listeners? Specifically, it was hypothesised that:

1. People whose PTCs have shifted tips would also have thresholds in TEN exceeding 10 dB of the TEN level/ERB;
2. People whose PTCs do not show shifted tips would also have thresholds in TEN within 10 dB of the TEN level/ERB.

**Method and Procedure:** Twenty-two normally hearing adults and fifty-three hearing-impaired adults with hearing loss ranging from mild to profound degrees participated in this study. All stimuli were presented monaurally via Sennheiser HD25 SP earphones.

**Psychophysical tuning curves**

The probe signals comprised of pure tones at 500, 1000, 2000, and 4000 Hz, which were pulsed on for 300 msec and off for the same duration, with rise-fall times of 10 msec. The maskers comprised of narrow-band noises produced by filtering white noise (the filter slopes were greater than 60 dB/octave). The centre frequencies of the noise maskers were 0.24, 0.43, 0.78, 0.92, 1.0, 1.08, and 1.23 times the signal frequency. The noise bandwidths were 80 Hz for probe signals at 500 and 1000 Hz, and one-third of the critical bandwidth for probe signals at 2000 and 4000 Hz (100 Hz and 230 Hz respectively). The stimuli were recorded on audio compact discs, with pulsed probe tones on one channel, and narrow-band noise maskers on the other channel. A broadband pink noise was added at 50 dB below the level of the masker to minimise the effect of detection of combination tones that might arise from interactions between the probe tone and the narrow band noise components on the tuning curve measurements.

When measuring tuning curves, the probe tone was presented at 10 dB above threshold (10 dB SL). The subject was asked to adjust the level of a narrow-band noise masker until the probe tone was just inaudible. The masker level was the averaged level of three to five adjustments. This procedure was repeated for each of the 7 noise maskers for each probe signal frequency. The probe tone was presented at 5 dB SL for 26 measurements due to limitations of loudness discomfort.

**Threshold Equalizing Noise (TEN) test**

The threshold-equalising noise as recorded on the compact disc supplied by Starkey was used. The pure tones were pulsed in the same way as the probe tones used for measuring psychophysical tuning curves as described previously. Test frequencies were 250, 500, 1000, 1500, 2000, 3000, 4000, 6000, and 8000 Hz.

The absolute threshold for a pure tone was first determined using the adjustment procedure as described above. The TEN noise was then presented at 70 dB/ERB (equivalent rectangular bandwidth), and the subject was asked to adjust the level of the tone until it was just audible in the presence of the noise. For people with more severe hearing loss, the TEN noise was presented at a level that was determined by the individual. The subject was asked to adjust a remotely controlled attenuator to find the noise level that could be tolerated without causing loudness discomfort. The noise masker was presented at a level rounded to the nearest 5 dB of the adjusted level. The maximum level at which the TEN noise was presented was 105 dB/ERB.
Analysis
The results from the PTC test were used to estimate the tip frequency for determining shifts, if any. Shifts, in octaves, were calculated by:

\[ \text{Shift}_{i} = \log_{2} \left( \frac{f_{\text{tip}}}{f_{\text{sig}}} \right) \]

where \( f_{\text{tip}} \) is the frequency of the masker with the lowest level and \( f_{\text{sig}} \) is the frequency of the signal. A shift value of zero denotes no shift, i.e., the signal was perceived in the cochlear region that was tuned to the signal frequency.

The sharpness of tuning (\( Q_{10} \)) was calculated by taking the bandwidth at 10 dB above the masker level at the signal frequency \( f_{\text{sig}} \), expressed as a ratio of the signal frequency using the formula:

\[ Q_{10} = \frac{f_{\text{sig}}}{\text{bandwidth}} \]

A \( Q_{10} \) of zero denotes that frequency selectivity is so poor, or that the tuning curve is so flat, that the bandwidth at 10 dB above the masker level at the signal frequency is infinitely wide.

Data from the TEN test were analysed by calculating the elevation, defined as follows:

Elevation, = Masked threshold, – max (absolute threshold, TEN level/ERB).

An elevation of greater than 10 dB indicates a dead region.

Results:

Audiometric thresholds
The mean absolute thresholds of the test ear of all subjects are shown in Figure 1.

Figure 2 shows the mean masked thresholds of 22 normal-hearing participants, measured using a TEN level of 70 dB/ERB. The error bars represent the 95% confidence range. The mean thresholds are all within 4 dB of the TEN level/ERB. On the upper side, the 95% confidence range falls within 7 dB of the TEN level/ERB for all test frequencies. These results are consistent with those reported by Moore (2000) and Moore et al (2003).

The results were categorized according to whether it was possible to measure absolute thresholds and whether the noise could be made sufficiently intense to provide 10 dB masking. Out of the ears that met these criteria, the number of positive diagnoses of dead regions was determined by considering...
whether the threshold in TEN exceeded 10 dB of the TEN level/ERB. Table 1 shows, for each of the octave frequencies 0.5, 1, 2 and 4 kHz, the number, out of 53 impaired ears, for which (1) an absolute threshold could be measured; (2) the TEN noise was sufficiently intense to give 10 dB masking; and (3) a positive diagnosis of dead region was obtained.

Table 1.

<table>
<thead>
<tr>
<th>Frequency (Hz)</th>
<th>Absolute threshold</th>
<th>10 dB masking (% out of 53 impaired ears)</th>
<th>Dead region</th>
</tr>
</thead>
<tbody>
<tr>
<td>500</td>
<td>53</td>
<td>48 (90.6 %)</td>
<td>1 (1.9 %)</td>
</tr>
<tr>
<td>1000</td>
<td>53</td>
<td>50 (94.3 %)</td>
<td>2 (3.8 %)</td>
</tr>
<tr>
<td>2000</td>
<td>53</td>
<td>39 (73.6 %)</td>
<td>6 (11.3 %)</td>
</tr>
<tr>
<td>4000</td>
<td>49</td>
<td>35 (66.0 %)</td>
<td>6 (11.3 %)</td>
</tr>
</tbody>
</table>

As shown, the TEN test did not give a diagnosis for 26 – 34% (2 – 4 kHz) of the impaired ears. These included cases where the hearing loss of an individual is so severe that an absolute threshold cannot be measured, and cases where the TEN cannot be made intense enough to give 10 dB masking. Among the hearing-impaired subjects for whom the masked threshold of the test tone in the TEN was within 10 dB of the absolute threshold, some of them had masked thresholds that were also within 10 dB of the TEN level/ERB (5 at 500 Hz, 8 at 1000 Hz, 3 at 2000 Hz, and 7 at 4000 Hz). However, it cannot be concluded that a dead region is not present. In other cases where the masked threshold was within 10 dB of the absolute threshold, it was above 10 dB of the TEN level/ERB (2 at 2000 Hz, and 4 at 4000 Hz). It is likely that a dead region may be present, but the result is inconclusive.

In the cases where the TEN noise could be made sufficiently intense to provide 10 dB of masking, the prevalence of dead regions was lower at lower frequencies (<4 % at 0.5 and 1 kHz) than at high frequencies (>10 % at 2 kHz and 4 kHz). These rates are much lower than those reported by Moore et al (2003), which indicated that 72% (8 out of 11 ears) had dead regions at 1 kHz; 61% (8 out of 13 ears) had dead regions at 2 kHz; and 71% (10 out of 14 ears) had dead regions at 4 kHz. A failure to meet the criteria for positive diagnosis of a dead region does not necessarily indicate absence of a dead region at a specific frequency, as many of the TEN test results are inconclusive. The present results are possibly an underestimation of the prevalence of cochlear dead regions due to the method used for diagnosis.

The way in which the elevation of threshold in the TEN varies with hearing threshold level is shown.
in Figure 3. Data points above the horizontal line at 10 dB represent ears diagnosed with dead regions. Subjects with PTC tips in place are denoted by circles, and those with PTC tips shifted in frequency are denoted by filled triangles.

The increase in signal level with increasing hearing loss is consistent with broadening of auditory bandwidth for people with more impaired hearing. Some ears that were diagnosed as having dead regions did not show shifted PTC tips (circles above the horizontal line at 10 dB), vice versa, there were ears with shifted tips that were not diagnosed as having dead regions (filled triangles below the horizontal line).

Table 2 summarises results from all subjects. Nil indicates that the test could not be conducted due to limits of loudness discomfort.

<table>
<thead>
<tr>
<th></th>
<th>Not dead</th>
<th>Dead</th>
<th>TEN Inconclusive</th>
<th>Nil</th>
<th>Total</th>
</tr>
</thead>
<tbody>
<tr>
<td>500 Hz PTC Tip in place</td>
<td>51</td>
<td>1</td>
<td>3</td>
<td>55</td>
<td></td>
</tr>
<tr>
<td>Tip shifted</td>
<td>18</td>
<td>0</td>
<td>2</td>
<td>20</td>
<td></td>
</tr>
<tr>
<td>Nil</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>1000 Hz PTC Tip in place</td>
<td>61</td>
<td>2</td>
<td>2</td>
<td>65</td>
<td></td>
</tr>
<tr>
<td>Tip shifted</td>
<td>9</td>
<td>0</td>
<td>1</td>
<td>10</td>
<td></td>
</tr>
<tr>
<td>Nil</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>2000 Hz PTC Tip in place</td>
<td>52</td>
<td>2</td>
<td>6</td>
<td>60</td>
<td></td>
</tr>
<tr>
<td>Tip shifted</td>
<td>3</td>
<td>3</td>
<td>3</td>
<td>11</td>
<td></td>
</tr>
<tr>
<td>Nil</td>
<td>1</td>
<td>1</td>
<td>2</td>
<td>4</td>
<td></td>
</tr>
<tr>
<td>4000 Hz PTC Tip in place</td>
<td>49</td>
<td>1</td>
<td>4</td>
<td>50</td>
<td></td>
</tr>
<tr>
<td>Tip shifted</td>
<td>7</td>
<td>4</td>
<td>2</td>
<td>14</td>
<td></td>
</tr>
<tr>
<td>Nil</td>
<td>1</td>
<td>4</td>
<td>6</td>
<td>11</td>
<td></td>
</tr>
<tr>
<td>Total</td>
<td>250</td>
<td>15</td>
<td>24</td>
<td>11</td>
<td>300</td>
</tr>
</tbody>
</table>

Table 2. Consistency between TEN and PTC results expressed in terms of number of ears tested. Results that agreed are in bold print.

Of the 15 instances where the TEN test diagnosed a dead region, only 7 showed PTC tips shifted in frequency. The remaining 6 had PTC tips in place, and 2 had absent PTC results. In 37 cases, the PTC results revealed tips shifted in frequency whereas the TEN test did not indicate a dead region. Twenty-four of the TEN measures were inconclusive, out of which 8 had shifted PTC tips and 11 had PTCs with tips in place.

Eleven subjects met the criteria for a dead region at some frequencies. Seven of them had sloping hearing loss. Figure 4 shows one of these ears (PS69). In the left panel, the solid horizontal line represents the TEN level/ERB used for measurement, and horizontal dashed line denotes 10 dB above the TEN level. Filled circles represent absolute thresholds, and open squares represent thresholds in TEN. Masked thresholds on or above the dashed line are indicative of dead regions. In the right panel, the PTCs for 0.5 (open squares), 1 (open diamonds), 2 kHz (open triangles) are shown. The level of the probe signal is marked by the asterisks.

For PS69, the dead regions appear to extend from 2 kHz upwards, although significant masking was not possible at the higher frequencies where the hearing loss was profound. It is likely that the dead region starts at 1.5 kHz, as the masked threshold is almost 10 dB above the TEN level/ERB (8.4 dB), and 16.7 dB above the absolute threshold at that frequency. Consistent with the TEN test result that did not show
dead regions below 1.5 kHz, the PTCs at 0.5 and 1 kHz did not show shifted tips. The PTC at 2 kHz showed a tip in place, based on levels of the maskers centred at 1 and 1.08 times. Other masker levels reached output limits of the equipment and could not be measured.

This preliminary analysis of PTC and TEN test results reveals a lack of agreement between the results of the two tests in diagnosing cochlear dead regions. This may be related to the criteria used in the tests. For instance, the absolute shifts in PTC tips were very small in many cases and might have been due to measurement error. Further, some of the PTCs with tips in place may have occurred due to the detection of beats and simple difference tones. This is especially so for some PTCs with distorted shapes. In a similar vein, although a dead region is not deemed to be present when the masked threshold of the test tone in the TEN was 10 dB or more above the absolute threshold, but was not 10 dB or more above the TEN level/ERB, there were instances when the masked threshold was close to 10 dB (8 or 9 dB) above the TEN level/ERB, and the presence of a dead region cannot be ruled out.

Significance:
Information about the presence of cochlear dead regions may be useful in prescribing hearing aids, and in explaining the likely benefits of amplification for speech intelligibility. The value of such information depends in part on whether cochlear dead regions can be reliably identified.

References


INTRODUCTION

This paper presents teachers' and children's views on sound-field amplification intervention in mainstream cross-cultural classroom. The information was obtained from questionnaires. A rating scale, Teacher Opinions re Performance in Classrooms (T.O.P.I.C.) was also developed to provide information on teacher perceptions regarding perceived changes in student performance in unamplified and amplified classrooms. The findings are presented.

METHOD

• 142 subjects: 43% Vietnamese, Samoan, Spanish and Aboriginal and Torres Strait Islander
  18% other varied ethnic backgrounds
  39% English backgrounds
• Eight teachers and eight classes of Grade 2 children
• No prior experience with sound-field amplification technology
• Design: Four classes with “normal” listening conditions for half the year and four classes with amplified listening conditions for half the year. Listening conditions alternated mid-year.

INSTRUMENTATION

TEACHER and CHILD QUESTIONNAIRES

Purpose:
To ascertain teachers’ and children’s opinions and attitudes in the following key areas:
• reactions to technology
• opinions regarding any perceived benefits
• opinions regarding any perceived limitations
• the extent of use of the single-channel and dual-channel transmission options

Examples:
Would you like to keep the systems? If yes, why? (teacher questionnaire)
Tell me what you liked about the teacher wearing the microphone. (child questionnaire)

INSTRUMENTATION (cont.)

T.O.P.I.C. RATING SCALE

Eight questions relating to teachers’ perceptions of student performance in areas of attention, communication and classroom behaviour.

Rating scale instructions: Please circle the number that best represents child’s behaviour:

Scale:
1 = poor performance
2 = no change in performance
3 = positive performance

CONTENT OF QUESTIONS:

ATTENTION:
QUES. 1. Distractibility of child
QUES. 2. Child’s response to teacher
QUES. 3. Child’s response to peers

COMMUNICATION:
QUES. 4. Changes in use of verbal communication between teacher and child
QUES. 5. Changes in use of verbal communication between child and peers
QUES. 6. Changes in use of non-verbal communication between child and peers

CLASSROOM BEHAVIOUR:
QUES. 7. Changes in child-initiated voluntary communication
QUES. 7. Changes in child’s confidence
TEACHER AND CHILD QUESTIONNAIRE RESULTS

**BENEFITS:**
- "ease of hearing", "improved listening" (teachers and children)
- "less vocal strain", "less fatigue at the end of the day"
- "my class was more attentive", "children participated more" (teachers)
- "we liked it louder and clearer", "we can hear people with little voices" (children)

**LIMITATIONS:**
- "technical interference at times" (teachers and children)
- "having second transmitter did not encourage children to project voices when speaking without the systems" (teachers)

**DAILY USE OF SYSTEMS:**
- 33% of teachers used systems for all classes
- 42% of teachers used systems for more than one hour but less than two hours
- 25% of teachers used systems for more than two hours

**DAILY USE OF DUAL-CHANNEL TRANSMISSION:**
- 63% of teachers used second microphone for up to one hour
- 37% of teachers used second microphone for up to 30 minutes

CONCLUSIONS AND OBSERVATIONS

**T.O.P.I.C. RATING SCALE**

Teachers observed significant improvement in attention, communication, and classroom behaviour.
Also, according to the teachers, sound-field amplification intervention……
- facilitated interaction with peers.
- increased children's verbal contribution to classroom discussion.
- encouraged the children to adopt a more proactive and confident role.

**TEACHER AND CHILD QUESTIONNAIRES**

- The provision of ongoing training and technical support to teachers' needs to be addressed.
- Group rather than individual in-service training may improve outcomes.
- Greater emphasis on microphone techniques and exploring alternative teaching strategies during in-service training to teachers.
- The length of time systems were used during a school day was disappointing.
- The children were positive towards using the systems and particularly enjoyed using the microphone for morning talks.
Introduction
The normal auditory system can use interaural differences in either speech or noise to improve intelligibility. These differences comprise
• interaural level differences, arising from head shadow; and
• interaural time differences, arising from the relative distance between the source and the two ears.

How much of the binaural advantage is due to use of interaural time differences?
Subjects: 5 normally hearing young adults.
Method: BKB sentences and speech-shaped noise were presented via headphones:
- monotonically to each ear (Mon),
- diotically to both ears simultaneously (Bin S0N0), and
- dichotically to both ears simultaneously, with noise delayed by 700µsec in the right ear (Bin S0N700-R) or the left ear (Bin S0N700-L).
The delay was chosen to approximate the human inter-ear delay time for laterally incident sounds.
For each test condition, the signal-to-noise ratio (SNR) for 50% correct was determined.
Results:
- Monaural SNR was not significantly different from binaural SNR without delay (p > 0.05).
- Binaural SNRs with interaural delay were significantly different from those obtained without interaural delay (p < 0.0003).
- Binaural SNRs with interaural delay introduced in different ears did not differ significantly between ears (p > 0.05).

Conclusion: Normally hearing adults demonstrated 3 dB improvement when information about normal interaural time difference was available.

Binaural advantages with bimodal hearing measured in the sound field:
Higher speech scores were obtained with cochlear implant and hearing aid (CIHA) than with cochlear implant alone (CI), both for children and adults. (Ching et al., 2001; 2003)
How much of the binaural advantage with bimodal hearing is due to use of interaural time differences?

**Subjects:** 5 children and 4 adults who normally used the Nucleus CI-22 or CI-24 system in one ear and a hearing aid in the opposite ear.

**Method:** BKB sentences and speech-shaped noise were presented via direct audio input under three conditions:
- monotonically to cochlear implant alone (CI),
- diotically to CI and HA simultaneously (CIHA – S\_0\_N\_0), and
- dichotically to CI and HA simultaneously - noise was delayed by 700µsec in the ear with a HA (CIHA-S\_0\_N\_700\_H) or in the ear with a CI (CIHA-S\_0\_N\_700\_C).

For each test condition, the signal-to-noise ratio (SNR) for 50% correct was determined.

**Results:**

- Performance of children was not significantly different from that of adults (p > 0.05).
- Monaural SNR was not significantly different from binaural SNR (p > 0.05).
- Binaural SNRs obtained with interaural delay were not significantly different from those obtained without delay (p >0.05).
- Binaural SNRs obtained with delay in the ear with a CI were not significantly different from those obtained with delay in the ear with a HA (p> 0.05).

**Conclusion:** Children and adults who used bimodal hearing could not make use of normal interaural time differences to improve speech perception in noise, regardless of whether the delay was in the ear with a cochlear implant or a hearing aid.

**Inferences:**

Either
- the hearing aid does not adequately preserve timing information, or
- the cochlear implant does not adequately preserve timing information, or
- the devices produce grossly different time delays, or
- the auditory system cannot combine timing information from the two modalities.

**Overall conclusions:**

- Normal hearing adults used interaural time differences in noise to improve speech intelligibility.
- Children and adults with bimodal hearing could not use interaural time differences provided by the two devices. Binaural advantage observed with bimodal hearing in the sound field is thus due mainly to one ear having a better signal-to-noise ratio arising from head-shadow effects, with some possible contribution from the complementary nature of the signals provided by the two devices.

Cooperative Research Centre for CI & HA Innovation, Australia.
Australian and international standards work

Hearing Aid EMC Standard – IEC 60118-13

Eric Burwood continued work with IEC (International Electrotechnical Commission) Technical Committee 29/Working Group 13. This working group is systematically updating the IEC 60118 series of hearing aid standards. The final draft international standard for the second edition of IEC 60118-13: “Electroacoustics – Hearing aids – Part 13: Electromagnetic compatibility (EMC)” has been circulated to National Committees for voting. This document has the status of a product EMC standard for hearing aids and at this stage only editorial changes can be made. In many respects this standard is similar to the Australian/New Zealand Standard AS/NZS 1088.9 Amendment No. 1 1996 for EMC requirements for hearing aids, however there are significant differences. The Australian/New Zealand standard is more stringent in that it protects hearing aids from interference when a bystander’s digital mobile telephone is more than one metre away from the hearing aid whereas the IEC standard uses a less stringent distance of 2 metres. Also when a hearing aid wearer is using a digital mobile telephone the Australian/New Zealand standard ensures that not more than 10% of hearing aid wearers using a digital mobile telephone may perceive interference under worst case conditions, whereas the IEC standard is less stringent where 50% of hearing aid wearers using digital mobile telephones may perceive interference under worst case conditions. The reason for choosing the less stringent criterion is based on the fact that digital mobile telephones are not used at all times in worst case reception situations. Hearing aids meeting the requirements of the Australian/New Zealand standard will provide hearing aid wearers interference free reception in more situations in both the bystander and mobile phone user modes than hearing aids that just meet the IEC standard.

Hearing Aid Loops – IEC 60118-4:

Eric Burwood has continued his involvement with the IEC (International Electrotechnical Commission) Technical Committee 29 / Maintenance Team 20 that has been reviewing IEC 60118-4: "Hearing aids. Part 4: Magnetic field strength in audio-frequency induction loops for hearing aid purposes". This standard is now over 30 years old. The proposed revisions are extensive and include different test signals such as recorded speech, simulated speech, pink noise as well as sinusoidal test signals. Correlating use of the different test signals has proved a much larger task than originally envisaged, but it reflects the different test methods used in different countries. The proposed title of the revised standard is "Induction loop systems for hearing aid purposes – magnetic field strength". The maintenance team has been working towards a Committee Draft of the revised standard. When finalised it will be circulated to the National Committees for a five-month voting period.

Design for Access and Mobility – AS/NZS 1428-5

Eric Burwood and Dr. Harvey Dillon have made significant contributions in the drafting of a new fifth part of AS1428: Design for Access and Mobility, titled “Communication for people who are deaf or hearing impaired” and it is intended to be used by planners, designers, regulators, builders and facility managers so that appropriate design solutions can be chosen to help hearing aid users and people who have varying degrees of deafness. The standard covers assistive listening systems, including loop systems, radio systems and infrared systems as well as environmental interferences in relation to both indoor and outdoor situations. Also covered are early warning systems including auditory, visual and tactile, and visual displays for public announcements and telephone services. The importance of adequate illumination is discussed and a guide to lighting levels is provided.

Occupational Noise Management – AS/NZS 1269

As Chairman of Standards Australia Committee AV/003 Acoustic – Human Effects, Warwick Williams oversaw the revision of AS/NZS 1269; 1998 Occupational noise management almost completed after significant response from the ‘public comment’ phase of the change process. Work on the revision is expected to be complete by the
end of 2004. Additions to the Standard have included greater detail on ototoxicity and more information on otoacoustic emissions.

**Narelle Murray** was also involved in the finalisation of AS/NZ 1269, particularly with the inclusion of an Informative Appendix in Part 4 (Auditory Assessment) on the use of Otoacoustic Emissions in Hearing Loss Prevention Programs.

**Toy standard – AS/NZ ISO 8124-1**

There has been considerable discussion of the current version of the Australian ‘toy’ (AS/NZS ISO 8124 – 1:2000) standard. Australia adopted the existing International Standard toy standard in 2000. The process of adopting existing International Standards is favoured by many sectors of industry and the Australian Government as Australian Standards that place performance requirements on devices that are not required by International Standards can be regarded as de facto trade barriers. The majority of Australian authorities would like to see the noise performance requirements of toys as mandatory and not optional as is now the case. **Warwick Williams** is assisting with input to this dilemma.

**ISO – general: Warwick** has continued to have significant input to International acoustic standards with participation at the ISO (International Organisation for Standardisation) working group meeting in Berlin, September, 2003. In particular there were significant contributions to the development of the uncertainty budget development for several Standards. This work is progressed on a continuous basis as well as during the meetings proper.


In 2001 the Australian Communications Industry Forum (ACIF) established an Acoustic Safety Working Committee to look at the acoustic safety of telephones and in particular, telephone headsets. The committee was established following concerns raised by the Australian Communications Authority and various industry bodies regarding an apparent increase in the number of telephone acoustic shock incidents. In particular, there were concerns about the acoustic safety of headsets used in the rapidly growing call-centre industry.

The CECRP/WC12 Acoustic Safety Working Committee was established under the auspices of the Customer Equipment and Cable Reference Panel (CECRP) of the ACIF. **Michael Fisher and Dr Harvey Dillon** have represented NAL on the committee and have performed research on behalf of the committee in the areas of acoustics, measurement, psychoacoustics and acoustic shock.

The committee concluded its work in 2004 with the issuing of two documents, the AS/ACIF S004:2004, Australian Standard: Voice frequency performance requirements for Customer Equipment and the ACIF G616:2004 Guideline Acoustic safety for telephone equipment. Revision of the S004 Standard included aligning the maximum sound pressure levels for handsets and headsets with the trend in international standards.

The G616 Guideline is a new document. It suggests generic steps and measures intended to help organisations, such as call-centres, reduce the risk and severity of acoustic shock injury. It recommends that equipment with sound pressure limits lower than those in the S004 Standard be employed in high-risk situations such as call-centres where staff, typically using headsets, are making or receiving a large number of calls involving different connections.

A maximum sound pressure level of 102 dBA SPL at the eardrum reference point is specified, although levels lower than this are recommended provided adequate speech intelligibility can still be obtained. The Guideline alerts the reader to a conflict between the low sound level required for protection and the higher sound level required for good speech understanding amidst background noise, and mentions methods, other than simple limiting, by which protection equipment can address this problem.
NAL Library

Joy Fischer

The major event for the library this year has been the move to new quarters within the Head Office building. Part of the old technology workshop area, for many years used as a store room, was refurbished to provide a pleasant light-filled library. The new library is smaller than the previous space but some rationalisation of resources and a different floor plan have combined to create a more efficient and welcoming area for staff and visitors. The trouble-free move was made possible by the co-operation of many staff, in particular, Rod Genford, Property Officer, and the IT team of John Seymour and Scott Brewer who kept the phones and computers operating. The new location, closer to the research area, has resulted in greater number of staff visits and greater utilisation of the resources. A collection of Bruel & Kjaer technical manuals, essential for the maintenance of much of the older equipment, has been transferred to the library and is being established as a special collection.

The library has been involved in the establishment of an Australian Hearing museum collection. This is an on-going project which aims to gather and catalogue equipment and artifacts of historical interest to Australian audiology and, in particular, Australian Hearing. Many items of interest have been collected including early audiometers, hearing aids, batteries and testing equipment. A room has been allocated on Level 2 for storage. Mosaic, a cataloguing software for small museum collections, has been purchased and loaded on a library terminal. Cataloguing of the collection should begin later this year.

Inter-library loan services continue to be the mainstay of the library services. The inter-library loan networks allow access for NAL staff to resources not held in house and also make our resources available to the wider Australian scientific community. NAL Library is a founding member of Gratisnet, a national consortium of medical libraries and also of Kinetica Document Delivery service, the national inter-library loan service managed by the National Library of Australia. Through these two networks we have access to all the major collections in academic libraries throughout Australia and New Zealand as well as many of the smaller hospital and medical libraries which have a wealth of resources. Our own very valuable backruns of journals are also made available to the wider scientific community in this way.

The library hosted a student from the Diploma in Library and Information Studies course Sydney Institute of Technology for 8 weeks in the second semester. The students in this course must do two industry placements as part of the diploma. The increasing number of one-person special libraries makes placement in the NAL library a valuable experience when seeking employment. Staff and students from Macquarie University Audiology courses continue to use the library throughout the year and several enquiries for information have been received by email from overseas students.

A corner of the relocated NAL Library.
### Projects approved by The Australian Hearing Human Research Ethics Committee

The following projects carried out at NAL or AH in 2003/2004 had the approval of The Australian Hearing Human Research Ethics Committee:

<table>
<thead>
<tr>
<th>Project no.</th>
<th>Description</th>
<th>Project leader</th>
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<tbody>
<tr>
<td><strong>Hearing loss assessment</strong></td>
<td></td>
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<tr>
<td>AS 00.1</td>
<td>Infant hearing screening by PAMR</td>
<td>S Purdy</td>
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<tr>
<td>AS01.1</td>
<td>Discriminative cortical auditory evoked potentials for diagnosis</td>
<td>M Sharma</td>
</tr>
<tr>
<td>CRC C3.5</td>
<td>Computer-aided assessment of children</td>
<td>T Ching</td>
</tr>
<tr>
<td>HLP98.2</td>
<td>Binaural assessment using OAEs</td>
<td>E LePage</td>
</tr>
<tr>
<td>AS02.1</td>
<td>Hearing aid effect: cross-cultural study of attitudes of Aboriginal and non-Aboriginal children toward peers with and without hearing aids</td>
<td>A Yonovitz</td>
</tr>
<tr>
<td>NA</td>
<td>Incidence and natural history of ‘ping’ tinnitus</td>
<td>E LePage</td>
</tr>
<tr>
<td>NA</td>
<td>Test of Virtual Spatial Hearing (TVSH): auditory processing disorder study</td>
<td>E LePage</td>
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</tbody>
</table>

| **Prevention of hearing loss** |  |  |
| Prev 00.1 | Effective training for noise reduction | N Murray |
| Prev 00.2 | Improving OHS training via OAEs | W Williams |
| Prev 02.1 | Personal stereo noise exposure | W Williams |
| Prev 03.1 | Barriers to occupational noise exposure reduction | W Williams |
| HLP92.1 | Longitudinal studies of noise-exposed individuals | N Murray |
| HLP98.1 | Methodology for fast assessment of risk | E LePage |
| NA | Hearing protector testing (commercial testing) | P Alway |

| **Hearing aids and rehabilitation devices** |  |  |
| CRC B1.1 | Trainable aid | G Keidser |
| CRC B1.2 | Determination of optimal characteristics | G Keidser |
| CRC B2 | Acoustically transparent hearing aid | H Dillon |
| CRC B3.4 | Noise sensitive adaptive amplification | M Fisher |
| CRC B6 | Impact of gain mis-match on localisation | G Keidser |
| Eng 00.1 | 1800 MHz GSM interference | E Burwood |
| Eng 00.2 | 800 MHz CDMA interference | E Burwood |
| SPONS 03.1 | confidential | L Carter |
| SPONS 03.4 | confidential | L Carter |

| **Rehabilitation procedures** |  |  |
| CRC C1.2 | Hearing aid fitting electrophysiological assessment (former CRC C3.6) | S Purdy/ M Golding |
| CRC C1.7 | Electrophysiological evaluation of bilateral cochlear implants | C Psarros |
| CRC C3.1 | Normalization of overall loudness | K Smeds |
| CRC C3.2 | Hearing aid fitting with cochlear implants | T Ching |
| CRC C3.7 | Functional assessment of children | T Ching |
| CRC C3.8 | Hearing aid prescription for children | T Ching |
| CRC C3.9 | Factors affecting speech intelligibility in hearing-impaired people (previously HA02.1) | T Ching |
| CRC C6.3 | Normative data collection (pilot study) – BKB/ DeVault Sentences | T Ching |
| HA 01.1 | Clinical evaluation of AH’s guidelines for fitting multiple memory hearing aids | G Keidser |
| HA 01.2 | Relationship between clinician style & client attitude & rehabilitative outcome | M Clapin |
| HA03.1 | Normative horizontal localisation data | G Keidser |
| HA 04.1 | Advantages of completely-in-the-canal hearing aids | S Carlile – U of Sydney |
| SPONS 02.2 | confidential | G Keidser |
| SPONS 04.1 | confidential | L Carter |
| SPONS 04.2 | confidential | L Carter |
| SPONS 04.4 | confidential | L Carter |
| NA | Prevention & repair of problems in talk with adults with acquired hearing impairment | Louise Skelt- PhD student, ANU |

Appendicies

Annual Report 2003-2004 73
Appendicies

Scientific publications

Published


Ching TYC (2003) Should I wear a hearing aid if I have a cochlear implant? *CICADA*.


AS/NZS 1269.4:2004 Occupational noise management, APPENDIX H -- OTOSTACOEMISSIONS (Informative) Using early warning properties of click-evoked otoacoustic emissions for application to hearing loss prevention (circulated for public comment)


Accepted


Agung KB, Purdy SC, Kitamura C. The Ling sound test revisited. *Australian and New Zealand Journal of Audiology*.


Keidser G, Brew C, Brewer S, Dillon H, Grant F, Storey L. The preferred response slopes and two-channel compression ratios in twenty different listening conditions by hearing-impaired and normal-hearing subjects and their relationship to the acoustic input.


Murray NM, LePage EL. Hearing in inmates in NSW prisons.

Murray NM, LePage EL. Ageing effect in click-evoked otoacoustic emissions and pure tone audiometry in groups with different types of noise-exposure. *Noise and Health*.


Williams W. Noise exposure levels from personal stereo use.

Williams W. Instruction and the improvement of hearing protector performance.

Williams W. A practical measure for workplace noise assessment and action.

Williams W. Self-reported hearing loss.


Smeds K, Keidser G, Zakis J, Dillon H, Leijon A, Grant F, Convery E, Brew C. Preferred overall loudness. II: Listening through hearing aids in the field and in the laboratory

### Conference Presentations


Appendicies


Keidser G. What the clinician should consider when selecting the amplification characteristic(s) in modern hearing aids. Paper presented at the Scandinavian Connexx Pro Workshop, Lillehammer, February 2004.


Other Public Talks

Carter L. Discussion on rehabilitation goals/technology at the Annual General Meeting of the Self Help For Hard of Hearing People, Chatswood Branch, March 2004.


Murray NM, LePage EL. Bringing coals to Newcastle. NSW Branch, Audiological Society of Australia, Newcastle, October 2003.


Appendicies

Visitors

Ms Paris Kostakos Advisor to Minister for Health & Ageing
Mr Tony Kingdom National Manager, Office of Hearing Services
Mr Geoff Plant Hearing Research Foundation, Boston, USA
Dr Jon Schallop Mayo Institute, Colorado, USA.
Dr Sally Lusk University of Michigan, USA
Prof John Bamford University of Manchester, UK
Dr Robert Morley Washington University, St Louis
Prof Simon Carlile Sydney University
Prof Philip Newall Macquarie University
Dr Robert Cowan Melbourne University
Prof Neville Fletcher Australian National University
Dr David Bies Independent researcher in South Australia
66 District Managers of AH Australian Hearing national centres
Mr Barry Roberts Siemens Hearing Instruments, Australia
Dr Stefan Launer Phonak AG, Switzerland
Dr Ole Dyrlund GN Resound, Denmark
Mr Stefan Bengtsson GN Otometrics, Denmark
Dr Elaine Saunders Dynamic Hearing, Melbourne
Dr Peter Blamey Dynamic Hearing, Melbourne
Mr Ken Miekl Acoustical Society of Australia
Dr Kikuo Aoki Association of Hearing Aid Research Promotion, Japan
Prof Jim Patrick Cochlear Ltd
Mr Peter Hanley Neuromonics
Ms Linda Laidlaw Neuromonics
Mr Jonathan Wolfe Vast Audio, Sydney
## NAL publications/materials

<table>
<thead>
<tr>
<th>Item No.</th>
<th>Description</th>
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<tbody>
<tr>
<td>P4376</td>
<td>NAL- NL1 Selection Procedure (Software and Manual)</td>
</tr>
<tr>
<td>P1266</td>
<td>Non-linear Hearing Aids and the NAL-NL1 Prescription Procedure (set of 2 VHS videos with notes)</td>
</tr>
<tr>
<td>P4071</td>
<td>NAL-OAE1 Software and Manual (NAL TEOAE Analysis Package)</td>
</tr>
<tr>
<td>P4083</td>
<td>Speech &amp; Noise CDs for Hearing Aid Evaluation – set of 3 CDs containing samples of speech and noise</td>
</tr>
<tr>
<td>P4380</td>
<td>Speech Recognition Materials on C D (AB Wordlists and NUCHIPS)</td>
</tr>
<tr>
<td>P4301</td>
<td>Manual of Speech Perception – containing instructions on clinical applications and score sheets</td>
</tr>
<tr>
<td>P4302</td>
<td>Speech Perception Test Package – containing items P4301, P4083 &amp; P4380, PBM’s &amp; PBN’s Test, and CID Everyday Sentences Test</td>
</tr>
<tr>
<td>P4381</td>
<td>COMMTRAM – A communication Training Program for Profoundly Deaf Adults</td>
</tr>
<tr>
<td>P4382</td>
<td>COMMTRAC – Modified Connected Discourse Tracking Exercises for Hearing Impaired Adults</td>
</tr>
<tr>
<td>P4384</td>
<td>Hearing Aid Selection Slide Rules (NAL-RP procedure)</td>
</tr>
<tr>
<td>P4385</td>
<td>PLOTT Test by Geoff Plant, 1983</td>
</tr>
<tr>
<td>P4386</td>
<td>The PLOTT Screening Test and The PLOTT Sentence Test, 1993</td>
</tr>
<tr>
<td>P4387</td>
<td>Percentage Loss of Hearing Tables (1982)</td>
</tr>
<tr>
<td>P4388</td>
<td>Improved Procedure for Determining % Loss of Hearing (NAL Report 118, 1988) and software for calculating PLH.</td>
</tr>
<tr>
<td>P4392</td>
<td>TACTAID II Training Program</td>
</tr>
<tr>
<td>P4393</td>
<td>Hearing Loss and Tinnitus Simulation CD</td>
</tr>
<tr>
<td>P4396</td>
<td>A Home Programme for Preschool Vibrotactile Aid Users</td>
</tr>
<tr>
<td>AC5825</td>
<td>Attenuation of Hearing Protectors 8th edition</td>
</tr>
<tr>
<td>P4377</td>
<td>“Damage Your Hearing &amp; It Won't Come Back” video</td>
</tr>
</tbody>
</table>

For further information about the above items or to order them, please visit our website: www.nal.gov.au

Or contact us directly -

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**Fax:** (+61) 2 9411 8273  
**E-mail:** Research@nal.gov.au
From Our Photo Album

Wendy Pearce and Maryanne Golding (back) review cortical evoked potential waveforms recorded from an infant with hearing aids.

Visiting scientist, Prof Wouter Dreschler, of AMC, KNO-Audiologie, investigates the accuracy with which hearing-impaired people can adjust their hearing aids in different situations.

Mandy Hill, Audiologist, demonstrating the use of functional evaluation as a tool for fine-tuning hearing aids for hearing-impaired children at the Parents’ Council for Deaf Education Annual Conference in Sydney.

Dr Sally Lusk, of University of Michigan, chats with Warwick Williams prior to giving a talk at NAL on hearing protector use.

Dr David Bies, an independent researcher in South Australia, shares his knowledge of the human ear and aerodynamic noise generation with NAL staff.

Nicola Schmitt, a student from Germany, working on the new BEST speech test materials during her 6-month work experience at NAL.

Dr Teresa Ching explains to Dr Kikuo Aoki of Association of Hearing Aid Research Promotion (right) and Paula Incerti from Cochlear Ltd. (left) about NAL research on the use of hearing aids with cochlear implants.

Emma Van Wanrooy, Audiologist, instructing Patrick on a loudness rating test.
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